THE ROLE OF PLANTAR FEEDBACK IN THE REGULATION OF HOPPING, WALKING AND GAIT INITIATION

By

PAUL FIOLKOWSKI

A DISSERTATION PRESENTED TO THE GRADUATE SCHOOL
OF THE UNIVERSITY OF FLORIDA IN PARTIAL FULFILLMENT
OF THE REQUIREMENTS FOR THE DEGREE OF
DOCTOR OF PHILOSOPHY

UNIVERSITY OF FLORIDA

2000

DEDICATION

Do'n lucht a chomhraic ar son an chirt ná dearmadfar a iobarthaí go deo.

Riobárd
Francis
Raymond
Patrick
Joseph
Martin
Kevin
Kieran
Thomas
Mick

Is feoilte caite 'tá na blátha scaipeadh ar do leaba chaoilse;
babhreá iad tamall ach thréig a dtaitneamh, nil snas ná brí iontu.
'S tá an bláth ba ghile liom dár fhás ar lithir riamh ná a fhásfaidh choíche
ag dreo sa talamh, is go deo ní thacfaidh ag cur éirí croí orm.

oHeageartaigh

ACKNOWLEDGMENTS

The spotted hawk swoops by and accuses me, he complains
of my gab and my loitering.
I too am not a bit tamed, I too am untranslatable,
I sound my barbaric yawp over the rooftops of the world.
Whitman

I would like to thank those people who stood by me when times seemed darkest, when everything was thrown into doubt and my future was clouded at best. I can not adequately express my gratitude for their support and encouragement through those times. In particular, there have been a few especially important people during this quest: Steve Dodd, (Ph.D.), despite the interruption to my coffee and bagel that fine June morning, turned my life in a new direction with one phone call; MaryBeth Horodyski's, (Ed.D.), patience, encouragement, and guidance prevented my falling by the wayside long ago, and she introduced me to the productivity of the predawn hours; and Denis Brunt, (Ed.D.), while providing a much needed tutelage in gait analysis, proved that even a Man' United fan can be cool.

Of all the acknowledgments that can be made regarding the attainment of this grail, a deeply felt thanks must go to my family, whose energy and actions have been a wellspring of strength in this endeavour.

All that is gold does not glitter,
Not all those who wander are lost,
The old that is strong does not wither,
Deep roots are not reached by the frost.
Tolkein

The strength we draw from our roots, which binds the clan, provides the energy to sustain us through turmoil, and the vigor to appreciate life, has been an invaluable ally. and perhaps now I'm done wandering, at least for a little while.

TABLE OF CONTENTS

Abstract of Dissertation Presented to the Graduate School
of the University of Florida in Partial Fulfillment of
the Requirements for the Degree of Doctor of Philosophy

THE ROLE OF PLANTAR FEEDBACK IN THE REGULATION OF
HOPPING, WALKING AND GAIT INITIATION

By

Paul Fiolkowski

December, 2000

Chairman: Stephen L. Dodd
Major Department: Exercise and Sport Sciences

In locomotion, the body is capable of adapting to a wider variety of terrain. It has also been noted that the leg can adapt to surfaces of different stiffness by altering leg stiffness. However, the mechanism by which the body adapts to different surfaces is not known. The purpose of this project to determine if plantar sensation plays a role in regulating locomotion in humans. There were three phases to this experiment, each examining a different aspect of locomotion.

The first experiment examined the changes in leg stiffness as feedback from the foot was diminished. Ten subjects (9 males and 1 female) participated in this experiment. Lidocaine was injected inferior and posterior to the lateral malleolus in order to achieve tibial nerve block at the level of the ankle. Tactile sensation, deep pressure sensation, and abductor hallucis activity displayed significant decreases following the injection, as did postural sway. Subjects demonstrated a significantly decreased leg stiffness after the

nerve block ($p<.001$). Total flexion of the leg increased, and there was later onset of quadriceps and gastrocnemius activation following the nerve block ($p<.001$).

Forward gait displayed fewer changes in kinematics, and no changes in muscle activation. Subjects demonstrated a shorter stride cycle ($p<.01$), a decreased step length ($p<.01$), and an increase in knee flexion at heelstrike ($p<.05$) in the anesthetized condition. In the third phase, gait initiation, there were no significant differences seen in any of the variables measured.

The results indicate that plantar feedback is important in regulating leg stiffness in a rapid rhythmic activity, as seen in the hopping, but has little impact in a slow precise movement such as gait initiation, which is apparently pre-programmed. The results also suggest that treadmill testing may not be a valid means of examining gait, as the treadmill may provide a drive to gait, masking possible differences between conditions

CHAPTER 1
THE ROLE OF PLANTAR FEEDBACK

Introduction

It has been observed previously that terrestrial locomotion involves moving along the ground in a bouncing fashion. This observation has been confirmed in a multitude of studies which have observed a plethora of bipedal and quadrapedal creatures (2, 10, 27, 49, 53, 65, 77, 88). When vertebrates walk or run, the center of mass rises and falls in a sinusoidal fashion as they move along the ground. In this way, the elastic energy in muscles tendons and ligaments is stored and returned with each step (11, 90, 124).

One advantage of using the legs like springs is that it minimizes energy expenditure at a given forward velocity (48, 49). The ability to act effectively as a system also requires the capacity to adapt. With regard to gait, an efficient system must be able to adapt to a variety of terrains. With regard to human locomotion, the ability to alter the leg spring as the surface changes is essential to maintaining an efficient gait.

The action of the leg as a linear spring in gait seems to have been firmly established (11, 26). Also, the leg spring is apparently capable of adapting to the varying surfaces it encounters (57).

Stiffness of a spring is defined as its deformation per unit force applied (N/m) and is denoted with the abbreviation k. A spring's stiffness is a measure of how much it will deform, or compress, when a load is applied to it. A stiffer spring will deform less than a

spring with a lower k value when a given load is applied. Work by Farley et al. (51) has indicated that the leg varies its stiffness in an inverse proportion to the stiffness of the surface on which it is acting, such that the overall system stiffness remains constant.

The ability of the body to sense changes in terrain and alter the parameters of gait seems an innate ability that will not only optimize energy expenditure, but also improve bipedal stability. As has been shown, a lack of stability is associated with an increased incidence of falls and injury (82). This has been shown in elderly as well as those with diabetic neuropathy (83).

In spite of the numerous observations that have been made concerning the regulation of gait and the mechanical properties of locomotion, there has not been a controlled examination as to the mechanism by which the body is able to vary the stiffness of the leg with regard to surface changes. The plantar surface of the foot is the only part of the body to come into contact with the ground during bipedal locomotion. It would seem logical that some aspect of the neurological functioning of the foot is responsible for transmission about the surface characteristics to the central nervous system (CNS), resulting in changes in leg stiffness.

This series of experiments was an attempt to determine the effect of sensory and muscular functions of the foot on the regulation of human locomotion. There were a total of 3 phases to this study, all measuring a different aspect of locomotion. All three experiments used blocking of the nerves supplying the plantar aspect of the foot as the independent variable. Part 1 was a study of a ballistic motion and the determination of leg stiffness in hopping. A force plate measured ground reaction forces, while kinematic data were recorded to determine joint angles and accelerations. The data were used to

calculate leg stiffness in the 2 conditions (i.e. anesthetized and normal). Part 2 examined forward gait using kinematic and EMG data. The last phase of the study, part 3, examined gait initiation, and the role that plantar feedback has in controlling the initiation of gait from a quiet standing position.

There were a number of kinetic and kinematic variables examined in these three phases, which will be further expanded upon in the following chapters. The information from this project could have implications for a number of areas. Perhaps foremost among these is the area of diabetic neuropathy. As these patients lose feedback from the foot, their risk for falls and injury increases dramatically (82, 123).

The role of afferent feedback from the foot in regulating gait is not yet understood, and it would be worthwhile to understand the contribution of the plantar surface sensory information to the control of gait. Another area in which this information may be useful is footwear design. As Robbins and his coworkers have repeatedly stated, the diminishing of plantar feedback via footwear may lead to altered kinematics that, over time, will contribute to an increased likelihood of injury (109, 110, 113, 114).

To date there have been no published attempts to ablate plantar feedback and subsequently evaluate dynamic activities. The information gained from this study can have implications for therapists, physicians, athletic trainers, and others treating injuries of the lower extremity, whether the interest is in rehabilitation of a previous injury or the development of a means of preventing future injuries.

Review Of Literature

There are a number of peripheral sensory receptors that contribute to postural and movement control (22, 31, 94). Muscle spindles transmit information about joint position based on the tension in the spindle (5, 22) A number of receptors in the joint capsules and ligaments relay information about tension in the joint, thus contributing to joint position sense and movement awareness (12, 30, 32, 69, 93, 107, 119, 126, 132). Mechanoreceptors of the skin can indicate pressure and load information, contributing to position sense by relaying information most likely associated with the location of the center of gravity (72, 87). The CNS is able to code this information spatially and temporally, thus creating an awareness of the position of the limb and its movements, along with an awareness of the position and orientation of the whole body (62).

Cutaneous receptors may be very important in postural control, although their role has been studied in a limited number of experiments (71, 74, 87). The receptors in the sole of the foot, which besides the palm of the hand is the only location of glabrous skin on the body, have been identified as Pacinian corpuscle-like, rapid adapting, slow adapting type I and slow adapting type II (87). It is an anatomical certainty that the plantar surface of the foot contains these receptors. The role of these receptors in the hand, with its structural similarity to the plantar surface of the foot, has been well-documented (43, 71, 127). The functional importance of the sensory structures in the sole of the foot has received only limited attention, with most of the previously published articles concerning postural stability (42, 87, 125).

One thing worth noting is that the receptors in the plantar surface of the foot have been assigned a variety of names by different authors: mechanoreceptors, pressor

receptors, load receptors and graviceptors (39, 74, 87, 93, 105, 115, 127). While it is not the purpose of this project to engage in a semantic duel over the proper allocation of a nominal description to these physiological units, the term mechanoreceptors will be used to refer to the sensory organs in the plantar surface of the foot. The use of the term is based on the their role in supplying sensor information to the CNS based on mechanical loads and deformations of the sole of the foot.

Postural Effects

The maintenance of upright posture has been the primary means of studying the role of plantar afferents. According to Kavounoudias et al. (74) the cutaneous mechanoreceptors of the plantar surface play an important role in maintaining balance. Noting that stimulation of different regions of the plantar surface elicited specific changes in postural equilibrium, the authors determined that the plantar surface is able to code into the CNS the location of pressure exerted against the plantar surface (73, 74). This certainly seems to fit in well with work by Clark et al. (31) and Ferrell and Smith (55) who noted that cutaneous mechanoreceptors provide facilatory input to the CNS that is used to interpret position and movement signals arising from other sources (31).

The significance of the plantar mechanoreceptors was also studied by Magnusson et al. (87) using hypothermia to reduce plantar feedback. The results indicated that information from the plantar surface is important in postural control. The authors also noted an interaction between plantar receptors and vision in the maintenance of postural control, an idea certainly supported by most models of postural control, in which there is

an integration of input from vision, labyrinth, muscle and joint proprioceptors and cutaneous receptors. A typical schema for this is shown in Figure 1-1.

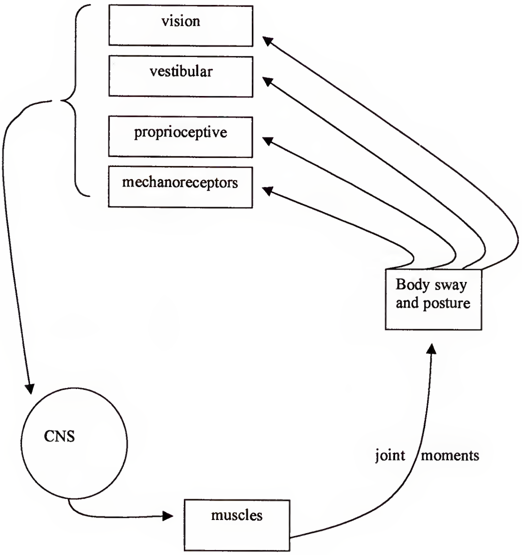

Figure 1-1. A schematic of human postural dynamics .

Control of Gait

When running and walking along the ground, humans use the stored energy is constantly stored and returned in the musculoskeletal system (47, 66). During gait, the complex systems of springs, composed of muscles, tendons, ligaments, and various connective tissues of the lower extremity behaves much like a single linear spring (89). The linearization of the components to form a simple spring has been previously revealed by Carter et al. (23) who demonstrated that stiffness has 3 components: passive, intrinsic, and reflex. They also noted that reflexes act to compensate for yield and linearize the mechanical properties of the limb, the components of which are inherently nonlinear (23). This has certainly supported other works in which the stance limb is modeled as a linear spring with a point mass equivalent to body mass (2, 11, 24, 91).

The stiffness of the leg spring will affect the dynamics of gait in several ways. Previous work has shown that kinematic variables, such as stride frequency and ground contact time, are affected by leg stiffness (90, 91). The stiffness of the limb, however, is not invariant, with vertebrates apparently able to adapt the stiffness of the limb to the terrain (1, 3, 54).

On any given day, people will encounter a wide variety of terrains during locomotion. These surfaces will likely have varying properties and compress more, or less, than others. The end result is essentially another spring in series with the spring-mass system of the person (54, 58). If human leg stiffness were indeed invariant, then efficiency would greatly decrease on a more compliant surface, as there would be a greater vertical oscillation of the center of mass with each stride cycle (48). There is recent work to indicate that humans are capable of adapting leg stiffness to a wide variety

of surface stiffness, as such that the overall stiffness of the leg-surface system remains nearly the same on either a hard or soft surface (51, 56-58).

The general equation for stiffness is

$$k = \mathrm{F}/\Delta \mathrm{L}$$

where k is the stiffness, F is the force, and ΔL is the deformation. The stiffness for the entire system, when considering gait on different surfaces, has been expressed as

$$k_{\text{total}} = k_{\text{leg}} + k_{\text{surface}}$$

We can calculate leg stiffness, assuming that the surface stiffness will not change over the course of the experiment. Stiffness is a function of deformation per unit force, and is expressed as Newtons per meter (N/m). Leg stiffness can be calculated by measuring the change in deformation of the leg relative to the force applied to it. Cavagna (24) demonstrated that leg stiffness can be determined from force plate measurements. By calculating the double integral of the vertical acceleration measurements, the change in position of the center of mass of the body can be determined. (24) Dividing this result into the peak force will give the stiffness of the leg spring. With the stiffness of the force plate as a known constant, the system stiffness can then be calculated.

What has not yet been determined, however, is the mechanism by which humans adapt their leg stiffness to the terrain. The works by Duysens et al. (45), and Robbins et al. (113) suggest that sensory aspects of the foot play a role in the regulation of the various aspects of gait (45, 113). The results of Magnusson et al. (87) clearly indicate that there is a role exerted by receptors of the foot in postural stability.

When the foot was hypothermic, a situation that would have decreased the transmission of impulses from the afferents in the sole of the foot, there was a significant increase in postural sway (87). Other authors have used ischemia to examine the effects of plantar afferents on proprioception (125). One unifying characteristic of these papers has been the steadfast use of static postural stability, as measured by postural sway on a force plate. We are unaware of any studies which have examined the role of plantar afferents during a dynamic task.

Lower Extremity

There is a tremendous complexity to the lower extremity, and it must perform almost contradictory functions in normal gait. In the initial part of the stance phase of gait, the leg must act as a shock absorber to absorb and disperse the energy from the impact of the foot with the ground. The foot and leg must then form a rigid lever to propel the body forward during the latter part of stance.

In the foot, especially the plantar aspect, there is a highly complex architecture that allows the foot to serve these varying roles. The arrangement of the bones, in supinated and pronated positions, the arrangement of connective tissues such as the plantar fascia, and the number of muscles in the foot, especially the plantar aspect, all serve to allow the foot to function in its varying capacities, and require an optimal integration of all structures to allow efficient gait and minimize injury incidence.

The coordination of these events relies on an efficient transmission of information from the various proprioceptors to ensure smooth muscle coordination. Similarly, there is, as previously stated, a need to adapt to the varying terrain and surfaces encountered

during gait (8, 54) This alteration of the leg stiffness, in inverse proportion to the surface stiffness, is possibly regulated by plantar feedback, which has not yet been demonstrated.

The coordination of the events of forward gait is heavily dependent on feedback from the various proprioceptors in the lower extremity. The afferent information from fast and slow acting receptors, tactile disks, Pacinian corpuscles, muscle spindles, joint capsule receptors, and Golgi tendon organs are coded temporally and spatially, with the resultant effect being modulation of locomotor tasks (13, 87, 118, 121). Each of these receptors plays a different role in the transmission of sensory information to the CNS. The specialization of the different organs allows each to impart a slightly different aspect of the limbs position, orientation, movement and rate of change of movement to be transmitted to the CNS (5, 78, 117). There is an interplay of higher processing and reflex activity that provide for adaptation to changing situations in dynamic activities.

There are a number of tactile receptors in the plantar region of the foot that transmit information to the CNS. Slowly adapting receptors (SAR) and fast adapting receptors (FAR) transmit information about cutaneous loads. Muscle spindles provide input to the CNS about the muscle length and rate of stretch that is being applied to a muscle (5, 130). It is likely the integration of all these receptors that provides the ability to maintain a stable and upright posture. As has been demonstrated, blocking of these afferents can decrease postural stability (125).

The majority of the plantar aspect of the foot is supplied by three peripheral nerves: the lateral plantar nerve, the medial tibial nerve, and the calcaneal nerve. These are all terminal branches from the tibial nerve, and serve both sensory and motor functions of the foot. The saphenous and sural nerves also supply some sensory functions

of the foot, on the extreme lateral and medial edges, respectively. The innervation of the lower extremity is shown in figure 1-2, while figure 1-3 depicts a more detailed branching of the tibial nerve into the medial and lateral plantar nerve to the foot.

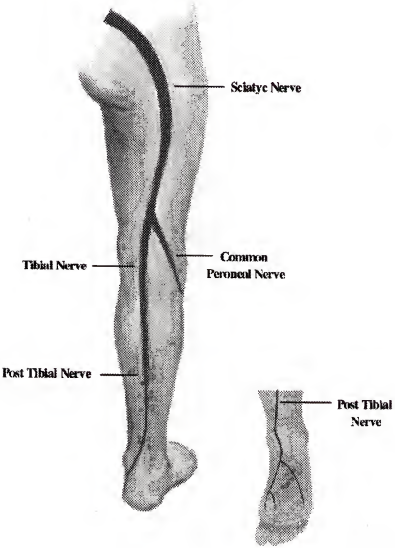

Figure 1-2. The nerve supply to the leg

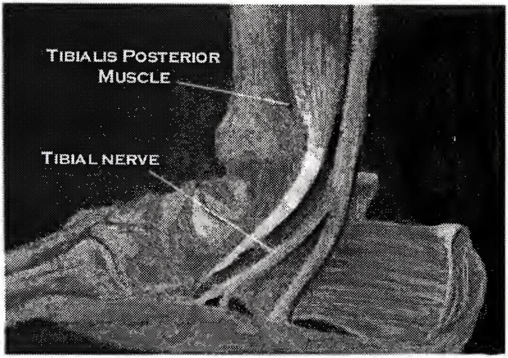

Figure 1-3. The branches of the tibial nerve to the foot.

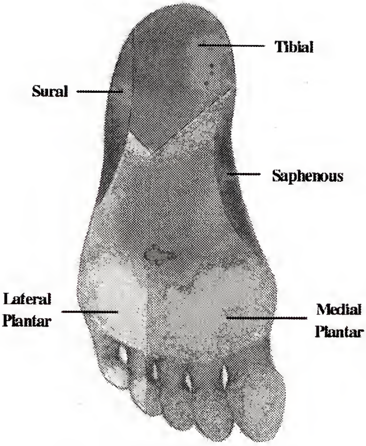

Figure 1-4. The sensory distribution to the plantar surface of the foot.

The role of sensory feedback from the foot is only recently gaining attention. One of the major reasons for this has been the awareness of peripheral neuropathy in elderly, especially in diabetic populations (81). The loss of plantar feedback has certainly been shown to lead to rather deleterious consequences to this patient population (28, 82). These individuals' insensate condition does not permit the detection of noxious stimuli, which would normally induce an alteration of stance.

Most of the published work in this area has been related to sensory loss in peripheral neuropathy. What has been examined far less are alterations in gait with peripheral neuropathy. It has been suggested that a loss of sensory input from the feet may influence the stability of gait (130). A recent publication by Dingwell et al. (41) has indicated that there is an increased variability in certain gait parameters of peripheral neuropathy patients (41). An increase in postural instability is believed to be associated with the loss of proprioceptive information (28, 130). With the data showing an adverse effect on postural stability with the loss of afferent feedback from the feet, it seems that this should extend to gait. The goal of this project will be to examine the role that feedback from the plantar region of the foot has on human locomotion.

CHAPTER 2
LEG STIFFNESS IN HOPPING

Introduction

The basic concept underlying the premise of leg stiffness is the spring-mass model of the leg in gait (11, 90). This essentially treats the leg as a simple linear spring, with the body treated as a point mass, equivalent to the mass of the entire body. During gait, this spring, comprised of the muscles, tendons, ligaments and other tissues compresses and rebounds with each step, storing and returning elastic energy (2, 89). An example of this concept is depicted in Figure 2-1. In this way, energy is conserved with each step, and the center of mass follows a predictable vertically oscillating path. (11, 25). In gait, the lower extremity behaves as a simple linear spring (89). The stiffness of the leg-surface system remains fairly constant over a wide range of surface stiffness levels (51). While the phenomenon has been observed and well reported, there has been no explanation for the basic mechanism by which people are able to adjust leg stiffness to a surface change. This stiffness has mostly been measured while hopping (47), but running studies have also been conducted, examining changes in leg stiffness, running velocity and stride frequency (50).

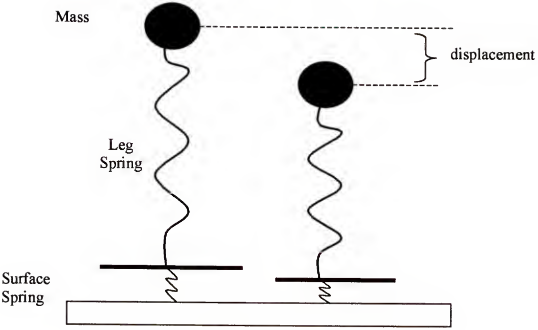

Figure 2-1. Depiction of springs in a hopping system.

With the published literature demonstrating an ability to adapt leg stiffness to the terrain, there have been limited attempts to account for the origin of the changes in leg stiffness. When the stiffness of the support system is altered, changes have been noted in the kinematics and kinetic of specific joints of the leg (7, 51). Farley et al. (51) have noted that there are measurable changes in the stiffness of the ankle joint when the activity surface is altered in human hopping. Specifically, these authors have identified the ankle as being responsible for the adaptations to changes in surface stiffness (51). The ankle displays changes in joint stiffness that correspond to increases in leg stiffness. These authors found no changes in knee or hip stiffness, which is in contrast to Arampatzis et al. (7), who noted significant changes in knee stiffness when running. With

the conflicting publications in this area, it remains to be seen how these changes in stiffness may arise. Farley et al. (51) have measured EMG of the lower extremity in an attempt to determine how the ankle stiffness is modulated, but their results noted no relationships between muscle activity and stiffness changes.

Another factor that has yet to be examined is the mechanism by which there is a signal to change muscle activity. One possibility is that plantar affect muscle activity, which in turn will affect leg stiffness. One published article has noted that in humans plantar nociceptive activity will affect tibialis anterior motoneurons (95). The results further indicated that the afferents from the tibial nerve exert a greater affect on tibialis anterior motoneurons than do the peroneal or sural nerve (95). The tibial nerve conducts efferent impulses to the muscles in the sole of the foot (via the medial and lateral plantar nerve) and carries afferent impulses from the sole of the foot. With its functional role clearly established, it seems reasonable to conclude that plantar afferents will affect motor activity.

There are a number of tactile receptors in the plantar surface of the foot that transmit information to the CNS. Slowly adapting receptors (SAR) and fast adapting receptors (FAR) transmit information about cutaneous loads. Muscle spindles provide input to the CNS about the rate of stretch that is being applied to a muscle (5, 130). It is likely the integration of all these receptors that provides the ability to maintain a stable and upright posture. As has been demonstrated, blocking of these afferents can decrease postural stability (125).

Load receptors have been generally divided into two main categories: main receptors and accessory receptors. The main receptors are true load receptors, the Golgi

tendon organs, and body support receptors, which include all cutaneous receptors on the sole of the feet. The accessory receptors are classified as neuromuscular receptors, i.e. the muscle spindles and joint receptors (44). The role of cutaneous load receptors has been shown to alter joint angles and affect loading (94). In a similar vein, it has been demonstrated that cutaneous receptors are capable of affecting muscle activity (71). In the upper extremity, the role of the cutaneous receptors has been documented thoroughly, with ample evidence to indicate that stimulation of the various cutaneous afferents is essential to the regulation of motor function of grasping muscles and control of grip strength and motor control of the arm and hand (71, 75, 86, 94, 104, 115, 124). If this is essential to the upper extremity, then cutaneous mechanoreceptors are probably a significant factor in regulating loads in gait.

The first part of this series of experiments was to measure changes in leg stiffness when plantar surface of the foot is anesthetized. Previous authors have demonstrated that leg stiffness is inversely related to the stiffness of the activity surface. When the leg and surface is examined as a whole system, the equation for the system has been defined as

$$K_{total} = k_{leg} + k_{surface}.$$

In this equation, k_{leg} = the stiffness of the leg, $k_{surface}$ is the stiffness of the activity surface, and k_{total} is the stiffness of the leg and surface as an entire system. The stiffness can be calculated using the following equation:

$$K = F/\Delta L.$$

The determination of the stiffness, k, is the result of dividing F (the peak vertical ground reaction force) by ΔL (the change in displacement of the center of mass).

The derivation of the numbers for the equation given was first demonstrated by Cavagna, et al. (24) Using a simple mathematical principle, it was shown that determining the displacement of the center of mass can be accomplished by calculating the second integral of the vertical acceleration. Integrating the acceleration will provide velocity, and integrating the velocity will provide the displacement (26). This has been a fundamental equation in the spring-mass models used in locomotion analysis (11, 36, 47, 100).

Stiffness Changes with Surfaces

It has been noted in recent years that the stiffness of the leg in running or hopping is not a constant. The stiffness has been shown to vary with forward speed, stride frequency and surface stiffness (50, 58, 90). Of these variables, surface stiffness is one that can not be controlled by the individual. Interestingly, there are adaptations exhibited by runners to different surfaces. Runners have been shown to alter a number of kinematics of gait, with the result being a change in leg stiffness that varies inversely with the stiffness of the surface (56). Variables that have been shown to change are the angular range traversed by the leg, foot contact time, and forward speed (49).

There has been no satisfactory explanation of the mechanisms by which these kinematics are varied. Ferris et al. (56-58) have shown that joint stiffness is varied in response to surface stiffness. In a hopping experiment it was demonstrated that it is ankle stiffness that is altered in order to produce the greatest changes in leg stiffness (51). In contrast, Arampatzis et al. (7) have published results showing that knee stiffness is the largest contributor to the leg stiffness.

Ferris et al. (56-58) has carried the work further, in examining the mechanisms responsible for the adjustments to surface stiffness. This was indicated by the transition from one surface to another, where subjects did not adapt leg stiffness in the first step when the stiffness change was unexpected. When the change was anticipated, the leg stiffness was adjusted prior to the fist step on the new surface.

If it is afferent feedback that regulates the stiffness, then plantar afferents are likely the mechanism behind the alterations. On a similarly basic level, it has not been answered as to whether it is muscle activation changes or other changes in kinematics that are causing leg stiffness to vary. Similarly, if muscle activation is a factor, how is the activity changing in response to surface changes?

These are basic questions that have yet to be addressed in the literature. More importantly, they point to a gap in the understanding of the basic regulation of gait, and the means by which the body adjusts to different conditions.

<u>Leg Stiffness</u>

The alteration of leg stiffness with changes in sensory feedback fits in well with the neuromechanical control hypothesis (60). In this hypothesis, it is emphasized that the role of reflexive neural feedback in a dynamic feedforward system will have an effect on the regulation of rapid rhythmic activity. By examining the role of afferent feedback and the effects on mechanical properties of the leg, i.e. stiffness, it may provide some measure of understanding the integration of neural and mechanical systems in legged locomotion.

One aspect of gait that has received attention in recent years is that of leg stiffness. This is a mechanical characteristic that is a function of force per unit change in

length. The stiffness (k) can be calculated from the peak vertical ground reaction force (F_z) divided by the distance the center of mass moves downward in stance (ΔL). The stiffness of the stance limb becomes more important as an adaptation to different surfaces. Figure 2-2 depicts the relationship between displacement and peak force in the determination of stiffness.

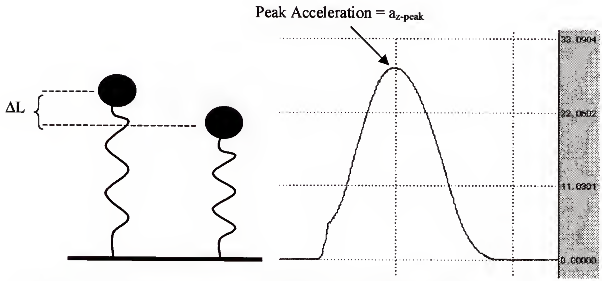

Figure 2-2. A depiction of the measures used in calculating leg stiffness.

ΔL = change in length of the leg spring
= $\iint a_z$ (double integral of the acceleration curve from ground contact to peak)

$F_{z\text{-peak}} = a_{z\text{-peak}} \bullet \text{mass}_{\text{subject}}$

k = stiffness
= $F_{z\text{-peak}} / \Delta L$

Methods

Participants

A total of 10 participants served as subjects for this study. All participants were free from lower extremity pathology and were in general good physical condition. All subjects were instructed in the data collection procedures, and then signed an Informed Consent, as approved by the University of Florida Institutional Review Board (IRB-01).

The testing took place in two conditions, with the independent variable being anesthetic condition. For this experiment, an orthopedic surgeon administered a subdermal injection of 1% lidocaine with epinephrine in the area of the tibial nerve, immediately posterior and distal to the medial malleolus. The amount was dependent on the weight of the subject, and varied between 3 and 10 cc, as determined by the surgeon.

Procedures

In order to evaluate the effect of lidocaine on the plantar tactile perception Semmes–Weinstein monofilaments were used to evaluate sensory deficits between pre- and post-injection status. The location of sites tested and the level of monofilament that is perceptible were recorded immediately before data collection, and then after the injection. Deep pressure detection was also recorded before and following the injection of the anesthetic. A Chatillon® pressure algometer was used to quantify the detection of deep pressure on the heel pad. The algometer head, a $2cm^2$ rubber circle, was pressed against the heel while the foot was manually held to prevent movement of the foot. The force (in lbs.) at which the subject was able to detect the pressure was then noted and recorded.

The effects of the anesthetic injection on plantar intrinsic musculature was recorded with surface EMG. A preamplified electrode was pressed against the surface of the skin over the abductor hallucis muscle, and a marker was used to note its location. The participant then performed a maximal voluntary isometric contraction while the muscle activity was recorded at 1000Hz. This was done in the control state and then in the anesthetized state. The data was then analyzed for the integrated value of each contraction, with the reference value being the control condition. While the amount of pressure holding the electrode against the skin may have varied, the differences were large enough to discount the possibility that the changes seen were the result of variance in electrode application.

Functional stability was assessed using a Romberg test. In this test, the subjects stood with their arms at their side, and then balanced on one leg. The force plates collected data at 1000 Hz in three planes (anteroposterior, mediolateral and vertical). The data was compared for the amplitude of the postural sway between conditions.

Small reflective markers were attached to the acromion, greater trochanter, lateral joint line of the knee, lateral malleolus and 5th metatarsophalangeal joint of the foot. Pre-amplified EMG electrodes were used to record muscle activity. Four electrodes were attached to the skin with double-sided adhesive tape. These were located over the rectus femoris, medial head of the gastrocnemius, medial soleus, and tibialis anterior.

For data collection, subjects hopped on a force plate at 2.2 Hz. This frequency has been shown that to be close to the natural frequency of hopping humans (56). All subjects were given as much time as necessary to practice the task and become accustomed to maintaining the rhythm. An electronic metronome maintained the rhythm with an audible

timer. At 2.2 Hz, the interval between vertical ground reaction force peaks should be 455 msec. Intervals between the peak ground reaction force that fell within 5% (±23 msec) were be considered as valid data to be used for comparisons. An example of data that was excluded is shown in figure 2-3 below. the time interval, as shown was 395 ms, and there was not a typical sinusoidal path to the COM in this period.

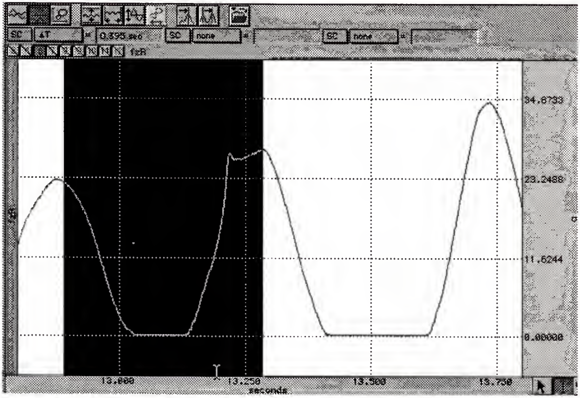

Figure 2-3. Example of force plate data that were considered unacceptable for calculation of leg stiffness.

Instrumentation

Force plate. An AMTI force plate was used for data collection. (American Mechanical Technologies, Inc., Watertown, Mass.). The force plate has a surface stiffness in the vertical direction of $10.5x10^8$ N/m. Data were collected using Acqknowledge® software and stored on a desktop computer.

Data were collected at 1000Hz. A band stop filter was used at 60hz to remove electrical noise from the collected data The data was analyzed for peak vertical acceleration (a_z), ground contact time, and center of mass displacement. The double integral of the Fz signal was used in the calculation of leg stiffness. Technical specifications for the force plate are given in Table 2-1.

Table 2-1. Specifications for the AMTI® force plate and amplifier

Vertical (F_z) load limit	10,000N
Horizontal (F_h) limit	4,000N
Surface stiffness (k_z)	$10.5 \text{ x}10^4$ kN/m
Inputs	6 four-arm Wheatstone bridges
Output	+/-10 volts into a minimum 2K ohm load
Gain	adjustable: 1000-4000, factory set at 4000
Filtering	selectable: 10.5Hz or 1050Hz 30-dB critically damped low-pass filter

EMG. The activities of 4 muscles (rectus femoris, gastrocnemius, medial soleus, and tibialis anterior) were recorded while the subjects hopped on a force plate. The data were collected at 1000 Hz through Acqknowledge® software and stored on a desktop computer for processing. The raw data were band pass filtered from 20 to 350 Hz, using

the Acqknowledge® filtering process, prior to full wave rectification. The bandpass filtering values were chosen after a Fast Fourier Transformation (FFT) of pilot data revealed that the bulk of the signal was contained between these points. An example of FFT results is shown in Figure 2-4.

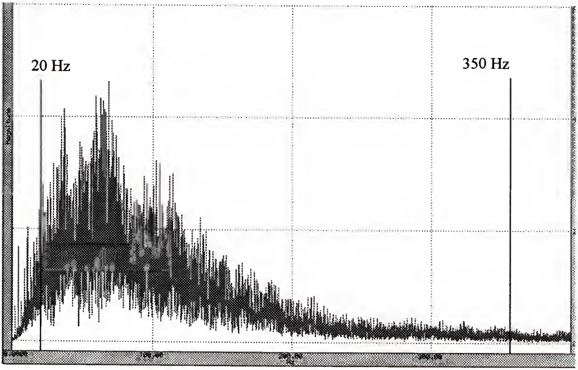

Figure 2-4. Fast Fourier transformation of an EMG signal to determine the cutoff frequencies for the bandpass filter.

The muscle activity was analyzed for time of muscle onset prior to ground contact, duration of muscle activity, maximal amplitude and mean amplitude. Onset of muscle activity was determined by the signal rising more than 2 standard deviations above the mean baseline level.

Kinematics. Data concerning the subjects' movements between the control and test conditions were collected using a MacReflex® system (Qualisys Inc., Glastonbury, CT.). Sagittal plane movements are the areas of concern in this project, as the leg joints contributing to stiffness in this task primarily move in this plane. Data were collected at 50Hz, using infrared light and a single camera, with retroreflective markers placed on the acromion, greater trochanter, lateral joint line of the knee, lateral malleolus and 5^{th} metatarsophalangeal joint. The technical specifications of the MacReflex® system are given in Table 2-2.

Table 2-2. Technical specifications for the MacReflex® measurement system.

Sampling rate	50Hz/camera
Exposure time	0.25ms
Field of View (FOV)	7-53 degrees
Resolution	1/30,000 of FOV
Spatio-temporal noise	0.003% of FOV
Temperature drift	0.01% of FOV over temperature range
Temperature range	0-40 degrees Celsius
Marker Size	0.5%-18% of FOV

Force plate and EMG data were collected using Acqknowledge software, while kinematic data was collected using MacReflex system (Qualisys, Inc., Glastonbury, CT). The kinematic data were exported into a spreadsheet for calculation of specified values. Kinematic data were compared for angles of the knee and ankle at touchdown as well as the angles for maximal flexion during the stance phase of hopping.

Results

The basic premise of this experiment was to examine changes in kinetics and kinematics when feedback from the plantar surface of the foot was ablated. In order to quantify the loss of feedback, we used three different measures to quantify the neural deficit: A Romberg test to quantify functional stability, monofilaments to test for light pressure sensation, and a pressure algometer to evaluate deep pressure sensation.

Plantar EMG

The results of the changes in EMG from the plantar surface, specifically the abductor hallucis muscle, were consistent among all subjects. The integrated EMG in the anesthetic condition averaged 22.53% of the control value. A depiction of a typical comparison of the EMG is shown in figure 2-5.

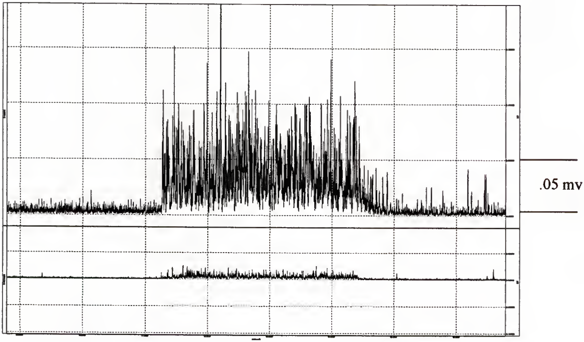

Figure 2-5. A typical EMG recording from the abductor hallucis. The top channels was recorded from a control trial while the bottom channel was taken from an anesthetized trial of the same subject.

Tactile Sensation

Measurement of tactile sensation was conducted using two different methods. Monofilament tests were used to assess light touch. A pressure algometer was used to evaluate the effects of the anesthetic on deep pressure sensation. The use of a pressure algometer for testing pressure sensation has been demonstrated by several authors, with their results indicating the method to be reliable and offer repeatable results (6, 79, 80, 99, 101, 106). The algometer was pressed against the sole of the foot in the heel region, as the foot was stabilized by another person to prevent the foot from moving as pressure was applied. When the subject was able to detect the pressure, the value was recorded. The results of the testing are shown in Table 3. The reselects of a paired t-test indicated that there was a significant ($p < .01$) increase in the threshold to detection of pressure.

Table 2-3. Changes in deep pressure sensation.

	control	anesthetic
mean	0	11.8409091*
s.d.	0	6.43260516
range		5 to 20+

*p<.01

Testing with monofilaments was done by recording the lowest perceptible filament pressed against the sole of the foot. Areas tested were the longitudinal arch and the toes., i.e. the non-callous parts of the foot, as the callous in the metatarsal heads and heel are not suitable for this type of testing. This type of testing has been previously established as having a high degree of reliability and repeatability (4, 9, 85, 102, 128). There was a remarkable consistency among the subjects with the monofilament testing, with the lowest detectable value being 3.61 kg filament. There was a significant difference in monofilament testing for tactile sensation between conditions. ($p<.01$)

Postural Stability

In order to determine the effects of the loss of feedback on functional measure, the Romberg test was used to measure postural stability in the mediolateral and anteroposterior directions. In order to quantify the relationships between posture and feedback loss, the amplitude of postural sway , as a percentage of body weight, was used to compare between conditions. Romberg testing following the injection increased the forces exerted in the mediolateral direction an average of 62% (range of increase: 26-215%). Forces exerted in the anteroposterior direction increased 52% (range of increase: 12-139%). All subjects who demonstrated a loss of deep pressure feedback also had the highest values for postural sway. An example of this is shown in figure 2-6 below.

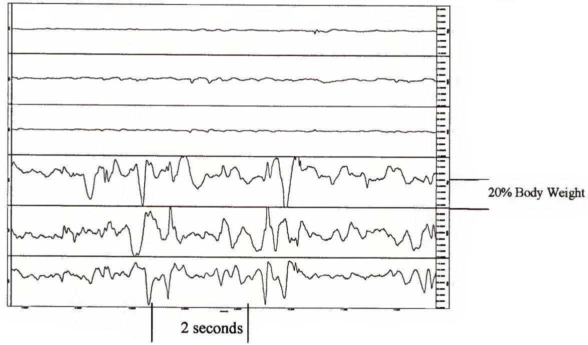

Figure 2-6. Force plate results from a single subject during the Romberg test. The top 3 channels (Fx, Fy and Fz respectively) were recorded in the control condition, while the bottom 3 channels were recorded after the nerve block.

Leg Stiffness

There was a significant decrease in the leg stiffness in the subjects as shown in figures 2-7. A repeated measures ANOVA revealed a significant decrease in leg stiffness between the 2 conditions (p<.001).

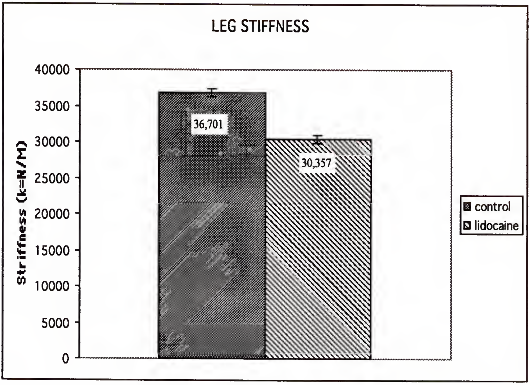

Figure 2-7. Comparison of leg stiffness between conditions(p<.01).

Joint Angles

Analysis of the joint angle at touchdown (TD) detected a significant difference in ankle angle (P<.05) yet no difference in knee angle was noted. The results are depicted in figure 2-8.

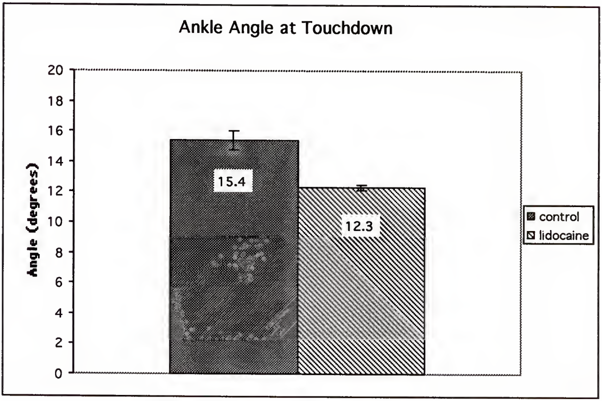

Figure 2-8. Ankle dorsiflexion at TD (p<.01)

Angular Displacement

The values for knee and ankle angle at the time of touchdown (TD) were compared between conditions for the subjects. However, it was noted that there was a difference in strategies between subjects when adapting to the loss of feedback. While there was a consistent change in leg stiffness, the same means of decreasing leg stiffness was not seen in the subjects: Some subjects demonstrated a greater knee flexion, but no difference in knee flexion was noted across conditions for all subjects. Some subjects exhibited a greater value for ankle flexion. Therefore, we combined the 2 values in order to calculate an overall flexion value. The difference between conditions is shown in figure 2-9. Total flexion in the control condition was 56.47°compared to 63.28° in the anesthetized condition ($p<.001$)

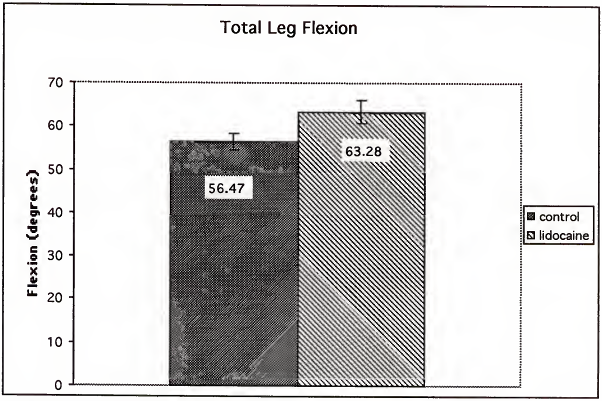

Figure 2-9. Comparison of total flexion, of the knee and ankle combined, between conditions. The difference is statistically significant ($p<.001$)

Muscle Activity

The muscle activity for the gastrocnemius and quadriceps were measured for this phase of the project. The dependent measures were muscle onset before touchdown and duration of muscle activation. Figure 2-10 depicts a typical comparison of the data collected, after full wave rectification of the EMG.

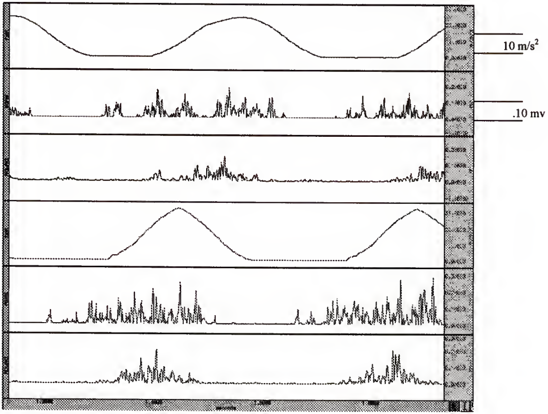

Figure 2-10. A display of the Fz data with gastrocnemius and quadriceps muscle activation. The bottom 3 channels are control trials.

Channel 1: vertical GRF - anesthetized
Channel 2: gastrocnemius - anesthetized
Channel 3: quadriceps - anesthetized
Channel 4: vertical GRF -control
Channel 5: gastrocnemius - control
Channel 6: quadriceps - control

Gastrocnemius. The gastrocnemius exhibited significant changes in timing between conditions, as is shown in figure 2-11. The muscle had a significantly later onset relative to touchdown in the anesthetized condition, when compared to the control condition ($p<.001$). There was no difference in muscle duration between conditions.

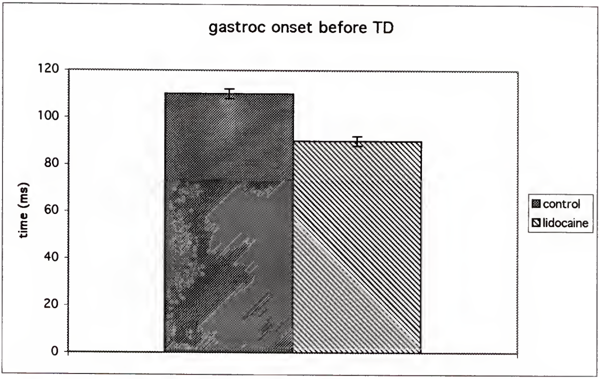

Figure 2-11. Overall comparison between conditions for onset of gastroc muscle activation prior to touchdown when hopping

Quadriceps. Similar to the gastrocnemius, the quadriceps demonstrated significant changes in timing between conditions ($p < 0.01$). The difference in muscle activation prior to touchdown is shown in figure 2-12. In the control condition, the quadriceps had an earlier activation, while the duration of activation demonstrated no difference between conditions.

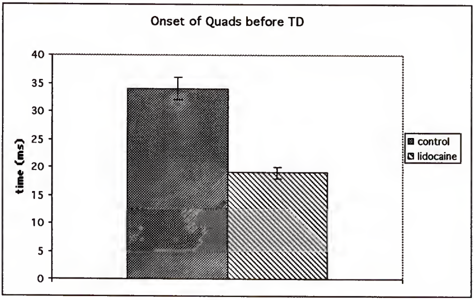

Figure 2-11. Quadriceps activation prior to touchdown in hopping ($p<.01$).

Discussion

Humans are capable of adjusting to a variety of conditions and speeds in locomotion. The regulation of the locomotor system allows for feedback to constantly adjust the system to optimize efficiency (1, 14). Mathematical models have done well in predicting the optimal compliance at given speeds for human gait (3). What has been lacking is the mechanism by which humans adjust to differing terrain in order to maintain this optimal condition. A notable absence in the majority of the literature is an integration of the neural and mechanical theories of locomotion.

The results of the experiments indicate that plantar feedback has a significant impact on regulating leg stiffness. Previous work has shown that humans adjust leg stiffness to changes in surface stiffness. In this way, the overall system stiffness, as indicated by the equation:

$$k_{\text{total}} = k_{\text{leg}} + k_{\text{surface}}$$

remained constant. (51, 56, 57). However, in this experiment, we were able to show that humans alter leg stiffness, with no change in surface stiffness. The loss of feedback from the sole of the foot resulted in a significant decrease in leg stiffness.

The increases in flexion of the joints of the leg are consistent with the decreases seen in leg stiffness. With the increase in postural instability, as assessed by the Romberg test, it is safe to state that plantar feedback is important in maintaining postural stability. These results are consistent with the published works of Magnusson (87) and Thoumie (125). These authors had previously demonstrated that patients with peripheral neuropathy exhibit a decreased postural stability. This experiment replicated similar results, indicating that the tibial block did induce a peripheral neuropathy for a short time.

It can be concluded from the results that the feedback from the sole of the foot is important in regulating load behavior in dynamic activities. The published results, supported by our tests, point to the role of plantar afferent feedback in maintaining postural stability. However, this is the first known attempt to determine if these receptors play a role in dynamic activities. The results of the kinetic data support the role of the plantar receptors in regulating load bearing behavior. What was surprising was the decrease in maximum force and stiffness in hopping. It had been supposed by previous authors that the loss of plantar feedback would lead to an increase in loading in gait (112). While diabetics with plantar neuropathy have shown an increase in the forces developed in gait, the opposite was shown here. The reason for this is not clear. With the sudden loss of feedback, the body may be demonstrating protective behavior, basically the subjects may have been subconsciously more cautious. However, they were able to maintain pace in hopping with the metronome, so we can discount the possibility that the subjects were slowing their activity.

The neural input for the foot seems to affect the properties of the spring-mass system, causing deviations from the equations derived by Blickhan (11). The equations for relating the basic functions:

$$my + ky = mg$$

seems to be valid, yet the equations derived from this are apparently not in agreement with the experimental data. A reasonable explanation for this may lie in the definition of the natural frequency, where:

$$\omega = k/m$$

and incorporating this value into the solution of the differential equation listed previously would result in

$$y = \dot{y}_a / \omega \sin \omega t - g / \omega^{2 \cos} \omega t + g\omega$$

where y = the deflection of the center of mass and $\dot{y}$ is the landing velocity. From this, it seems that the natural frequency of the system needs to be altered for the results to be consistent with the concept of a purely mechanically driven spring mass model of the locomotor system. The natural frequency may be changing in response to the lack of peripheral sensory input, and is an area that may need to be reconciled to the mechanical hypotheses for legged locomotion.

There are a number of approaches to take in examining the alteration in leg stiffness. Leg stiffness is going to be comprised of the stiffness of the joints of the knee, ankle and hip. Previous authors have stated that leg stiffness is primarily dependent on ankle stiffness (51, 52). Mathematical models were able to make accurate predictions based on changes in the geometry of specific joints of the body (51, 52). The stiffness of a joint will be dependent on its touchdown angle, the ROM of the joint, the activation of the muscles crossing the joint, and the condition of all the soft tissues surrounding the joint. In this experiment, we were able to observe changes in joint angles at touchdown, total ROM changes, and changes in muscle activation with the loss of plantar feedback. Our results differed from the previous authors in that there appear to be individual strategies for reacting to the deafferentation. While some subjects underwent a greater flexion at the knee during the landing phase, others experienced a greater ROM at the ankle. this is likely a reflection of different strategies used by the subjects to compensate for the loss of feedback.

One consistent factor noted among all subjects was the changes in quadriceps activation in the anesthetized condition: There was a later onset of muscle activity

relative to touchdown as well as a shorter duration. The changes in gastrocnemius activation was not as consistent among the subjects. A muscle's stiffness will depend on its level of activation and gain (96) and the decreased activation seen in this experiment would certainly support a decreased muscle stiffness, and a corresponding decrease in joint stiffness. This activation of a muscle before loading has been termed preflexive activity, which, during rapid activity when response time is shortened, can simplify control of the musculoskeletal system, allowing rejection of rapid perturbations (18). Whether the stiffness alteration is seen at the ankle or knee remains in question, but as the gastrocnemius crosses both the knee and ankle, it is reasonable to conclude that changes in the muscle's activation will affect both joints. In a similar vein of thought, alteration of the quadriceps will affect the knee as well as the hip, unless one joint is stabilized.

Regarding the innervation of the foot, the tibial nerve is a mixed nerve, and will affect both motor and sensory functions to the foot. We did note that there was a dramatic and significant decrease in activity of the intrinsic musculature of the foot, as measured by the abductor hallucis. While it was noted that all subjects exhibited a decrement in this activity, such decrement did not always correspond to alterations in the kinetics measured. The numbers do not support as conclusive claim to a preferential role of pressure receptors over muscle afferents in the regulation of leg stiffness, and the ability to specifically ablate one nerve or the other has more to do with anatomical variation , i.e. the bifurcation of the tibial nerve. Yet this does provide some further insight into the role of the nervous system as it affects the mechanical properties of the leg.

The results point to a dynamic coupling of the neural system with the mechanical properties of the leg and locomotor system. The neuromuscular mechanism responsible

for the changes seen is linked, apparently, to the feedback from the plantar surface of the foot. While the role in static posture had been demonstrated, we now have empirical evidence to point to the neural regulation of mechanical properties in rapid dynamic activities.

CHAPTER 3
THE ROLE OF PLANTAR FEEDBACK IN FORWARD GAIT

Introduction

From psychiatric conditions to physiological disease progression, whether kinetics, kinematics, inverse dynamics, feed forward modeling , there has been a vast plethora of published research in this field. It should also be noted that gait analyses have been conducted on nearly every imaginable species, from force vector analysis on cockroaches running over jello to locomotion energetics of African elephants (59, 61). The question then must be, can any further useful information be added to such a heavily worked vein of knowledge?

The answer to that lies in a question. That is, in order to provide further understanding to any area, one has to ask the right questions. Asking the right questions depends upon the ability to find a gap in the field of knowledge (S. Dodd, personal communication) The goal of the second part of this project was to determine what effects if any, that plantar feedback has on the kinetics and kinematics of forward gait.

The previous phase of this series of experiments studied the kinetics and kinematics of a simple dynamic task, hopping in place. In such a situation, the leg can be modeled as a simple linear spring attached to a point mass equivalent to the weight of the body. In forward gait, there are similarities to hopping. The center of mass of the body

moves in a sinusoidal fashion and the leg acts as a linear spring, in order to maximize efficiency and minimize energy expenditure (3, 11). This observation has been made in number of terrestrial species (10, 53, 91). In bipedal locomotion, the mechanical properties of the legs affects the vertical oscillations of the center of mass in forward gait As stiffness increases, the vertical oscillations decrease, with the inverse occurring with an increase in compliance of the legs (89, 91).

It also needs to be noted that in forward gait, there is a difference between running and walking. In running, there is always an aerial phase, where neither foot is on the ground (27). In walking, there is a double support phase, where both feet are on the ground (63). There is a similar contrast to the path of travel of the center of mass: In running the center of mass is at it's lowest point in mid-stance, while in walking it reaches a vertical apex at mid-stance (63).

The role of plantar feedback in the regulation of gait is an area that has been little studied in nearly a century of gait analysis. In spite of the knowledge that neuropathic conditions can have serious implications for diabetics and the elderly, (28, 41) there has been little effort to elucidate whether plantar feedback does have an effect on dynamic activities, and what the contribution to movement is. This phase of the study examined kinematics of the body when walking on a treadmill in normal condition and when the sole of the foot is anesthetized.

Review of Literature

In forward gait, there are a number of aspects that come into play in propulsion of the body. For stability and efficiency, the body depends on the central nervous system (CNS) processing and integrating a vast amount of information. Input comes from vision, labyrinth, joint and ligament receptors, cutaneous receptors and muscle spindles, among others (31, 62, 70, 73, 87, 118).

The ability of the body to sense changes in position and movement of its various parts is an important aspect to smooth movement and coordination of muscle activity in order to maintain an efficient gait (40, 45, 62). Previous literature has examined changes in proprioception when some aspect of feedback has been blocked. Changes in postural stability has been noted with the administration of treatments intended to block afferent input to the CNS (125). Decrements in the ability to detect passive motion as well as joint repositioning have been seen with neurological impairment, whether induced or the result of pathology (81, 87, 123).

Dynamic activities such as forward gait have been studied in patients exhibiting diabetic neuropathy (41). The main focus of such studies seems to be the pressures developing on the plantar surface o the foot (28, 81). As diabetic ulceration can have very deleterious consequences, this has certainly been a worthwhile endeavor. Indeed, such patients often exhibit an increase in localized pressure in static conditions (35). There is, however, little research that has examined kinematics of such patients in forward gait. Dingwell et al. (41) have shown an increase in variability of parameters of gait in neuropathic patients implying that this increased variability may indicate an instability in these people. Other studies have examined the incidence of falls and injury

in individuals with sensory deficits in the plantar aspect of the foot (83, 130). The authors have indicated that a decrement in this area does correlate with an increased risk of falls and injury.

In a line of research which has focused on the sensory capacity of the foot, Robbins et al. (108-114) have conducted a number of studies, the results of which seem to support the idea that the plantar surface of the foot is instrumental in proprioception and stability. One publication noted that there is a load sensing ability of the receptors of the foot that attempts to minimize plantar loads through protective reflexes (112). Another study noted that there is an increase in tone of the plantar intrinsic muscles in barefoot populations (111). The same authors also determined that a loss of plantar feedback, such as accompanies cushioned footwear, is associated with a decrease in position awareness, and that this decrement became more pronounced with elderly subjects (113, 114).

What has been distinctive in the literature is the paucity of observations of dynamic activities when feedback has been affected. McNair (92) noted that when footwear was changed, there was a change in the kinematics in running, as indicated by knee flexion through the gait cycle. More recently, Dewitt et al. (37) noted significant differences in kinematics and kinetics wen comparing barefoot running to shod running. Notably, there was a change in the foot angle at heel strike, and joint angular kinematics were different between conditions (37, 92).

What was lacking from these projects was an examination of the swing phase kinematics. It seems that in an effort to understand the forces experienced in running, there has been a focus, to the exclusion of all else, on the stance phase of gait (8, 76, 98,

124). It seems a logical extension that the body prepares for an impact with the ground by positioning the leg and foot during the swing phase of gait. The impact of the previous stride may result in changes for the subsequent support phase. Alternately, as has been suggested in a recent publication, there may be a crossed pathway from the contralateral leg in the support phase that modulates kinematics of the swing leg (95, 116).

Thus, the focus of the second part of this series of experiments was to examine the kinematics of gait following an ablatement of plantar afferent feedback. It is the hypothesis that plantar feedback will have a role in regulating forward gait. In the swing phase of gait, the leg is positioned to accept the load of the body, starting with the impact of the ground at heel strike. As locomotion is essentially a series of collisions with the ground, it is logical to suggest that the individual will attempt to optimize conditions for locomotion. Terrestrial bipeds need to maintain efficiency (53) yet keep loads below an injurious threshold (97) while maintaining stability (131) It is the intent here to determine what, if any, plantar feedback contributes to these goals.

Methods

Participants

A total of 10 participants, the same individuals who had participated in the first part of this project, served as subjects for this study. All participants were free from lower extremity pathology and were in general good physical condition. After signing an Informed Consent, subjects were instructed in the data collection.

The testing took place in two conditions, with the independent variable being anesthetic condition. For this experiment, and orthopedic surgeon administered a

subdermal injection of 1% lidocaine with epinephrine in the area of the tibial nerve, immediately posterior and distal to the medial malleolus. The amount was dependent on the weight of the subject, and varied between 3 and 10 cc, as determined by the surgeon.

Procedures

In order to evaluate the effect of lidocaine on the plantar tactile perception Semmes–Weinstein monofilaments were used to evaluate sensory deficits between pre- and post-injection status. The location of sites tested and the level of monofilament that is perceptible were recorded immediately before data collection, and then after the injection. Deep pressure detection was also recorded before and following the injection of the anesthetic. A Chatillon pressure algometer was used to quantify the detection of deep pressure on the heel pad. The algometer head, a $2cm^2$ rubber circle, was pressed against the heel while the foot was manually held to prevent movement of the foot. The force (in lbs.) at which the subject was able to detect the pressure was then noted and recorded.

Data Collection

Kinematic analysis. Small reflective markers were attached to the acromion, greater trochanter, lateral joint line of the knee, lateral malleolus and 5^{th} metatarsophalangeal joint of the foot. Data were collected over 30 second intervals while the subject walks on the treadmill.

Dependent variables The variables in this phase of the study were the linear and angular kinematics of the lower extremity. Evaluation of gait on a treadmill can present difficulties in detecting differences between conditions. Previous research has indicated that differences are masked by the treadmill, as the treadmill drives the gait by providing

neural input to the central patter generators (41). While the absolute differences can be expected to be minimal, if evident at all, there may be an increase in the coefficient of variation of the dependent variables.

A major difference between this project and the publication just cited is the dependent variables that were examined. If the feedback from the sole of the foot impacts kinetics in gait, it is likely to assume that the preparation to landing will be important in determining the forces exerted on the body. Therefore it is our intent to compare swing phase kinematics especially during the terminal swing phase, i.e. preceding heel strike. Statistical significance was set at p=0.05, with an ANOVA used to compare the variables between conditions and subjects.

Results

Kinematics

Timing of phases There were significant differences noted in the time duration of the phases of gait. In the anesthetized condition, there was a significant decrease on the stance phase of gait, with a corresponding decrease in the stride length as well as the total time taken for one stride. When feedback was ablated, the subjects reduced stride length while on the treadmill. Basically, the subjects took shorter, quicker steps on the treadmill in the anesthetized condition. There was, however, no difference in the stance time as a percentage of the total step time.

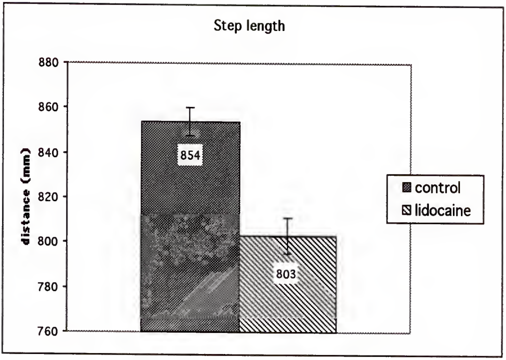

Figure 3-1 Comparison of the step length between conditions ($p<.001$).

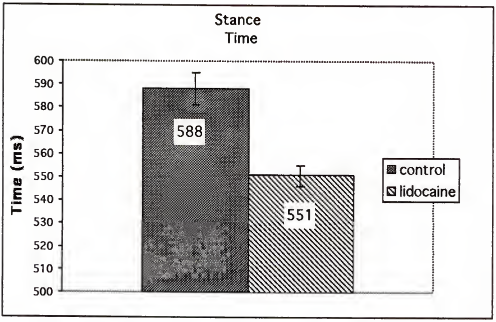

Figure 3-2. Stance time in gait compared between conditions ($p<.001$)

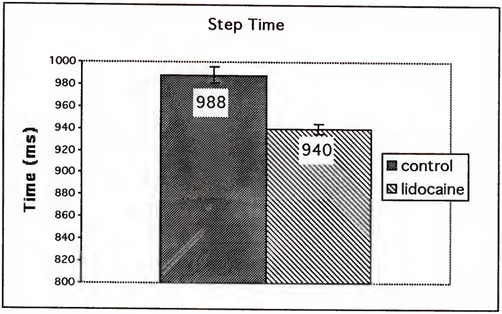

Figure 3-3 Step time of subjects between conditions
($p<.001$)

Joint kinematics There were no significant differences noted in the angle of the ankle at heelstrike, while there was a significant difference in the angle of the knee at heelstrike ($p<.001$) as depicted in figure 3-4. Subjects exhibited a more flexed knee at heelstrike in the anesthetic condition, compared to the control condition.

EMG

Due to technical difficulties, it was not possible to synchronize the kinematic data with the EMG data. Therefore, we were limited to comparing the duration and amplitude of the muscle data between conditions, without regard to its timing during the stride cycle. There were no significant changes seen in the gastrocnemius during gait on the treadmill. the muscle was analyzed for duration, mean amplitude and maximal amplitude.

No differences were noted between the control and anesthetized conditions. Similarly, there was no detected difference in muscle activity in the quadriceps between conditions.

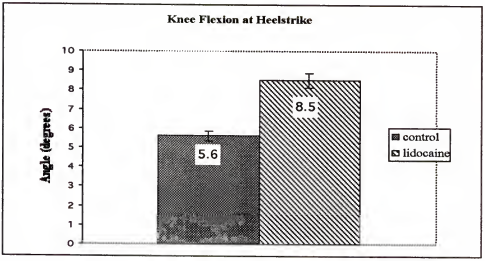

Figure 3-4. Comparison of knee angle at heelstrike between conditions. (p<.001)

Discussion

Compared to the number of measurable changes seen in the first experiment of this project, the treadmill data reflects few changes related to the loss of plantar feedback. The changes in kinematics were notable, yet the changes in EMG which were expected were not apparent in the muscles analyzed. Plantar nerves have been previously demonstrated to be involved in flexion reflex transmission (33, 84, 116). It had originally been hypothesized that the removal of plantar feedback would have resulted in measurable responses. The changes in joint angle at heelstrike were opposite from the results noted for the hopping phase of this project. While in hopping, a significant difference in ankle angle at touchdown was noted between conditions, this phase found a

difference in knee angle at heelstrike. The disparate results can likely be attributed to the tasks. One author had noted that in hopping, the ankle joint was primarily responsible for modulating leg stiffness (52). These results seem to support that concept. However, if these same changes in stiffness follow through with forward gait, one would have expected to note similar reactions in joint kinematics. While in hopping the ankle displayed a change in angle at touchdown, which fit in well with the model developed by Farley et al. (52), in forward gait, it was the angle of the knee at heelstrike which displayed a significant difference between conditions, with the anesthetized trials demonstrating an increase in knee flexion. If it is assumed a decrease in stiffness would be present in forward gait, as was noted in hopping, then the change in knee flexion is more consistent with the work of Arampatzis, who attributed changes in leg stiffness during gait to changes occurring at the knee (7).

The use of a treadmill for evaluation of gait changes was a likely factor in the lack of measurable differences between conditions, in spite of the subjective reports of the subjects and the objective measurements (the Romberg tests pre- and post-injection) that noted significant changes in postural stability and hopping.

With regard to the use of a treadmill, Sherrington noted that locomotion was possible in spinal cats (120). It was noted that on a treadmill, kinematics and muscle activity were remarkably similar, even though descending pathways had been entirely removed (122). Essentially, the work demonstrated that, at least in the case of feline quadrapeds, the treadmill essentially drives gait (46, 129). More recently, it has been seen that treadmill training can significantly improve the recovery of locomotor skills (68).

Treadmill training, in conjunction with body weight support, is used to retrain the muscle activation patterns and motor coordination of the muscles of the lower extremity (129).

Gait kinematics did undergo detectable changes between conditions; however, these changes were not entirely consistent with the changes noted in the hopping phase of this series of experiments. The subjects decreased their stride time, and decreased their step length, although the stance time as a percentage of stride time remained unchanged. This may have reflected some instability, with the subjects taking shorter steps after the nerve block. Previously reported data noted a decreased step length in elderly subjects compared to young subjects. The findings here are also consistent with the findings by Winter (133) who noted an increase on the coefficient of variation (CV) in gait kinematics. We did note an increase in CV in step length in the control trial of 11% while this rose to 18% in the anesthetized trial.

One aspect that seemed surprising in the data analysis was the lack of detectable changes in muscle activity. There was no significant in the gastrocnemius or quadriceps muscles, two muscles which are phasically active during the gait cycle. In the first part of this experiment it was noted that the timing of muscle onset varied between experimental conditions, presumably contributing to the altered leg stiffness. Activity of these two muscles, as indicated EMG recordings, showed no difference in duration of activation, nor in mean or maximal amplitude. A factor to consider in the lack of differences in muscle activation is the equipment used in data collection. Unlike the first phase of this series of experiments, there was no means of synchronizing the ground contact and EMG data. One of the significant findings in the first part of this project was that there was a later onset of muscle activity in the anesthetized condition. Whether or not that was true

for the treadmill trials is an area that could be explored, but there is a necessity for a more complex assemblage of equipment than is currently available to the investigators in this project.

The lack of difference in muscle activation between conditions may have been reflective of the effects of the treadmill on locomotion. While a significant difference in muscle activation was noted in the first phase of these experiments, no changes were noted during treadmill walking. Recent research into gait training for spinal cord injured (SCI) patients has focused on the use of the treadmill for gait retraining (38, 68). Much of this has built on the research begun by Sherrington, whose feline work has been previously noted (120). Essentially, it has been noted that the treadmill provides a drive to the CNS, resulting in muscle activation and EMG patterns that are not entirely under the control of higher CNS (46).

In addition to the drive supplied to the CNS by the motion of the treadmill, locomotion may be 'hardwired' into mammalian structures. If we assume that locomotion is a basic function, necessary for survival, e.g. eating and reproduction, then it may be a preprogrammed function, in a similar way that cardiac control is a basic neural function of mammalian physiology (64).

In summary, this experiment did not point to a definite role of plantar feedback providing a role in regulating forward gait. The role of plantar feedback may be to modulate locomotion, and improve the adaptation to terrain, or perturbations in gait. The treadmill provides a steady, forward task, at a fixed speed, over a monotonously even surface. If the role of peripheral feedback is to modify movement, then a more difficult task would be appropriate to test such a theory.

CHAPTER 4
PLANTAR FEEDBACK AND ITS EFFECT ON GAIT INITIATION

Gait initiation is considered the transition from a motionless standing position to the maintenance of steady state locomotion (20). This is certainly one of the more common locomotor tasks undertaken countless times on any given day. The task, however, requires a complex interaction of muscle synergy in order to provide the propulsive forward force as well as shift from a balanced bipedal state to a unipedal state.

The task of initiation of forward gait is normally undertaken with a desired direction and speed of motion in mind (16). Before initiation of forward gait can occur, there are a number of postural adjustments that need to happen, primarily aimed at shifting the weight distribution from both legs to one leg, the stance limb. Also, there is a change in the center of pressure between the limbs in both anteroposterior and mediolateral directions (17). The controlling factors in this task are still being studied, as well as the neurophysiological mechanisms responsible for the initiation of forward gait. It is the intent of this experiment to examine the role that plantar afferents, both the plantar mechanoreceptors and those associated with the intrinsic plantar musculature, have in contributing to the initiation of forward gait.

Review of Literature

The progression from a static upright posture to forward locomotion involves a transfer of weight to the support limb along with the simultaneous generation of a forward propulsive force (16). There is also a pattern of anticipatory postural adjustments that precede the initiation of forward movement (17). In human posture, numerous inputs from visual, vestibular and proprioceptive systems are responsible for maintenance of postural stability (74).

There is little information on the role of the proprioceptive system, specifically the feedback from the feet, in the process of gait initiation. Because they are at the interface between the body and ground, it seems reasonable to expect the mechanoreceptors to play a role in gait initiation (74).

One aspect of gait initiation that has been well described previously is the shift in the center of pressure in the initiation of forward movement (17). Following this line of reasoning, it seems that the ability to detect the center of pressure would be essential in controlling this process. The work by Magnusson et al. (87) indicated that pressor input from the foot exerts a significant effect in anterior-posterior postural control. Using hypothermic condition localized to the feet, the authors noted an increase in body sway, suggesting a decrease in postural control when the afferent signals from the feet were diminished, as would be expected with ice application (87). It would certainly seem logical that a decrement in anterior-posterior control of posture would have an adverse effect on gait initiation.

The role of plantar feedback in the regulation of gait also fits in well with the model posited by Full et al. (60). They suggest that during slow, variable frequency

locomotion tasks, the nervous system likely dominates by way of continuous feedback, with negligible effects of the mechanical system (60). A locomotion task that fits into the description, such as gait initiation, in which the tasks and sequences are pre-programmed, are likely to be little effected by feedback from the extremities.

Such a model can be useful, conceptually in linking the mechanical aspects of gait initiation with the neural input. The action of the individual in gait initiation has been described in detail, with the movement being approximated by the half-period of an inverted pendulum oscillating freely (15). These authors have asserted that the process of gait initiation is entirely dependent on the subjects' biomechanical constraints i.e. mass, inertia and position of the center of gravity (CG) with respect to the ground. The flaw, however, lies in the necessary use of the center of gravity.

The relationship between the CG and the center of pressure (CP) has been succinctly described by Winter (133) who defined the CP as essentially the neuromuscular response to perturbations or imbalances in the body's CG (133). If the CP is dependent on CG location, and CP determination by the body is a function of plantar mechanoreceptors, then it seems logical to postulate that gait initiation will be affected by the loss of this sensory feedback, and the biomechanical constants of the subject is insufficient to determine the temporal aspects of gait initiation.

Methods

Participants

A total of 10 participants served as subjects for this study. All participants were free from lower extremity pathology and were in general good physical condition. After signing an Informed Consent, subjects were instructed in the data collection.

The testing took place in two conditions, with the independent variable being anesthetic condition. For this experiment, an orthopedic surgeon administered a subdermal injection of 1% lidocaine with epinephrine in the area of the tibial nerve, immediately posterior and distal to the medial malleolus. The amount was dependent on the weight of the subject, and be between 5 and 10 cc, as determined by the surgeon.

Gait Initiation

For data collection, subjects stood with one leg placed on either force plate. The positions of their feet on the force plate were determined by markers on the force plate, so that each participant stood in the same position for each trial. Gait initiation was begun when the subject saw a small light come on. They then started walking forward at a normal pace.

Instrumentation

An AMTI force plate was used for data collection (American Mechanical Technologies, Inc., Watertown, MA). The force plate has a surface stiffness in the vertical direction of 10.5×10^7 N/m. Data were collected at 1000Hz using Acqknowledge® software on a desktop computer. A band stop filter was used at 60hz to remove electrical noise from the collected data

Force plate data were measured as a percentage of body weight (%BW). Comparison between anesthetized conditions were made for total time of gait initiation (the time from the first detected motion to the toe off of the stance limb).

Muscle Activity

EMG data were collected using pre-amplified silver-silver chloride electrodes (Therapeutics Unlimited, Iowa City, IA), an analog to digital converter and Acqknowledge® software (Biopac Systems, Goleta, CA). Kinetic data were collected from two force plates, measuring forces in the vertical (F_z), anteroposterior (F_x) and mediolateral planes (F_y). Static stability was measured using a Romberg test while standing on the force plate. EMG data were recorded from 6 muscles in the legs: Right gastrocnemius, left and right soleus, left and right tibialis anterior and right quadriceps. The EMG signals were collected at 1000Hz. The raw data were band-pass filtered from 20 to 350 HZ prior to full wave rectification of the signal. Onset of muscle activity was determined by the electrical signal rising more than 2 standard deviations above the mean baseline level and remaining at that level for 30 ms. Post processing of the data was done to measure muscle activity for time duration, maximal and mean muscle activity, and the integral of the muscle activity.

Data Analysis

The variables for muscle activity examined in this phase of the project were the timing of the soleus and tibialis anterior on both the swing leg and the stance leg. The force plate data were compared for the peak forces in the mediolateral, anteroposterior and vertical planes as well as the rate of AP force development. The values were entered into a spreadsheet, and analyzed for differences using Staview® statistical software (SAS

Institute, Cary, NC). The data were compared using a repeated measures ANOVA to test for effects between conditions. The level of significance for determining differences was set at $p = 0.05$.

Results

As has been described previously in the results in chapter 2, the measure of postural stability, the Romberg test, showed an increase in coefficient of variation (CV) in both the AP and mediolateral directions. This would indicate that there was a significant decrease in postural stability between conditions. A comparison of the time taken to complete gait initiation, measured as the time from the first measurable change in forces to the time the stance foot leaves the force plate, showed no difference. Figure 4-1 shows a typical gait initiation task as recorded from the force plates and selected muscles.

However, when the data were normalized as a percentage of the task, the relative timing of the phases of GI remained unchanged. the duration of TA activity was unchanged in the swing limb. There was a slight trend toward an increase in the stance limb TA duration of 11.6% ($p=0.095$). Time of onset, duration and mean activity of the soleus was also unchanged between conditions.

Kinetic profiles were also measured in this phase of the project. Similar to the muscle activities, there were no significant differences between the conditions for peak anteroposterior forces, rate of force development, and peak mediolateral forces. There were no significant change in mediolateral forces or anteroposterior forces under the stance limb in GI. The mediolateral and anteroposterior forces from one subject, averaged over 5 trials, are depicted in Figures 4-2 and 4-3, respectively.

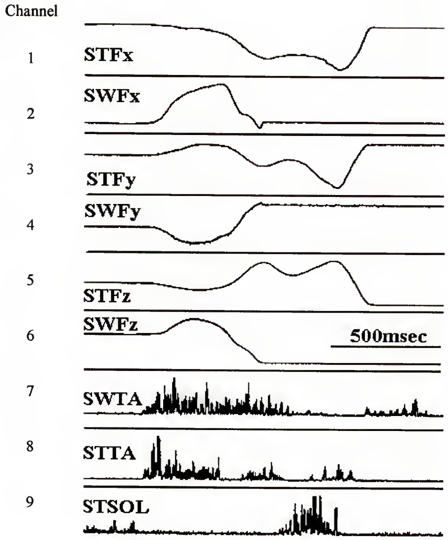

Figure 4-1. Typical data from gait initiation

channel 1	STFx:	stance anteroposterior forces
channel 2	SWFx:	swing anteroposterior forces
channel 3	STFy:	stance mediolateral forces
channel 4	SWFy:	swing mediolateral forces
channel 5	STFz:	stance vertical force
channel 6	SWFz:	swing vertical force
channel 7	SWTA:	swing tibialis anterior muscle
channel 8	STTA:	stance tibialis anterior muscle
channel 9	STSOL:	stance soleus muscle

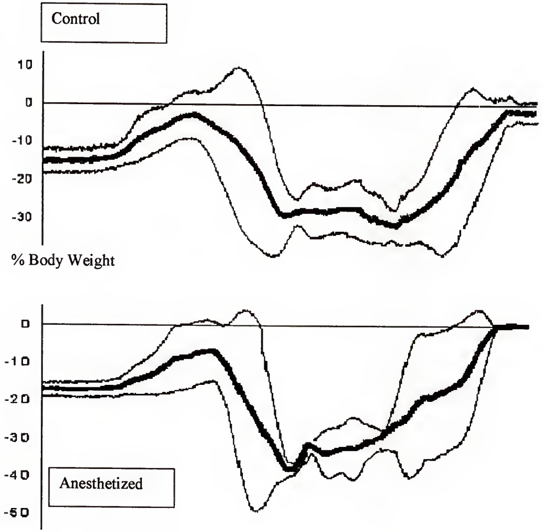

Figure 4-2. Mediolateral forces for one subject for 5 trials of gait initiation. The heavy line is the mean value, in %body weight, with the lighter lines denoting the standard deviation.

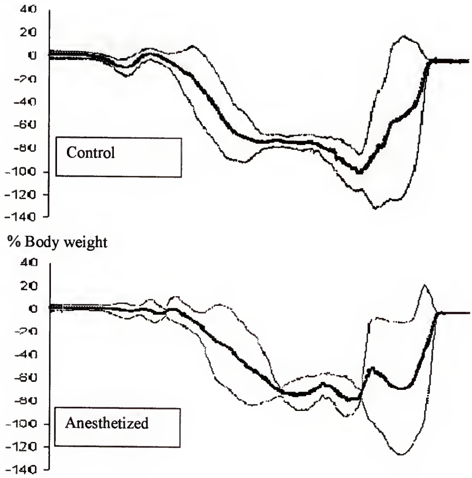

Figure 4-3. Anteroposterior forces averaged for one subject, showing the mean and standard deviation.

Discussion

A motor program may be defined as a structured set of central commands that define a temporal relationship of muscle activation to satisfy a goal oriented task.(34) The implications of these findings certainly seem to highlight the pre-programmed nature of the gait initiation (GI) task. In previous studies of GI, it has been determined that characteristics of the swing phase remain invariant, yet the stance phase, being the second part of the task, can be modulated depending on external conditions or the specificity of the task (16, 21). Our results, with the induced postural instability, demonstrated that both the swing phase and stance phase of GI are invariant.

With the significant changes in postural stability, it would have been expected to observe some change in GI. This should have been most apparent in the unilateral support phase of the task. However, as the results point out, this was not the case.

The reasoning for this may lie in the nature of the task. Gait initiation is a task that is performed countless times each day, and is essential to normal locomotion. Previous work has indicated that the task of gait initiation is a centrally programmed task, and thus will be relatively invariant (19). Similarly, one previously published work demonstrated that when sensory feedback was removed, temporal parameters of gait initiation showed no change (67). This certainly supports the concept that gait initiation is a centrally programmed task.

The lack of changes in the time to complete the tasks indicate that the initiation of gait, while apparently a centrally programmed task, is independent of sensory influence. Here, the instability and sensory ablatement may be seen as having an effect on GI. Separate studies have noted that foot position awareness decreases with age (108,

114), that there are changes in GI with age (29, 103) and decreases in postural stability with aging (105, 130).

Taken together with the previous literature ion this topic, the results of this experiment support the inextricable linking of postural stability, proprioception and GI. The data certainly support the previous reports that GI is apparently an invariant task. The initiation of gait is a centrally programmed task, yet plantar feedback does modify the task to some extent, as shown by the increased time to complete the task. The mechanism by which the afferent feedback from the plantar surface of the foot contributes to modification of tasks is an area that should be more fully investigated in future research.

CHAPTER 5
CONCLUSIONS

Summary

The data presented in this series of experiments is the first to systematically manipulate the level of plantar feedback in an effort to determine the effect of this specific aspect of proprioception on human locomotion. The results have certainly been mixed, with the range of effects of the sensory ablatement ranging from distinct differences in almost every measured area, as in the hopping experiments, to minimal detectable differences seen in the gait initiation. Three experiments were completed in this project in an attempt to determine the role that sensory feedback from the sole of the foot has on locomotion.

Part I: Leg stiffness in hopping

Humans are capable of varying their leg stiffness, and it has been shown that they can do this in response to changes in terrain. What has not been defined is the origin of signal that initiates the adaptability of leg stiffness to the surface. This phase of the project used a tibial nerve block to reduce plantar feedback and then measure changes in movement, specifically hopping.

The results indicated that leg stiffness is greatly affected by a loss of plantar feedback. Following a loss of intrinsic plantar muscle activity and a concomitant loss of

tactile feedback and deep pressure sensation, the subjects displayed a significant increase in postural instability, as measured by postural sway on the force plate. When hopping, the onset of muscle activation in the gastrocnemius and quadriceps occurred significantly later ($p<.01$), while flexion at the knee and ankle also increased ($p<.05$), resulting in a significant decrease in leg stiffness ($p<.01$).

Part 2: Forward gait

It had been expected that the changes seen in hopping, which is considered a control analogue to running, would persist into the forward gait phase of this project. Kinematics of gait did change between the conditions, with the subjects' step time, stance time and step distance all decreasing in the anesthetized condition ($p<.01$). There was a significant increase in knee flexion at heelstrike ($p<.05$), yet there were no detectable differences in EMG values between conditions. While the lack of measurable differences may have been unexpected, the use of a treadmill for forward gait may have provided significant neural input to the locomotor system, effectively masking any changes that may have occurred from the loss of plantar feedback.

Part 3: Gait initiation

Gait initiation is a common task in locomotion and is considered a preprogrammed task. There was no change in the time necessary to complete the task. After normalizing muscle activation times as a percentage of the task time, there were no differences in any other measured values for gait initiation. The results are in agreement with the tenet that GI is a pre-programmed task, with the results of this project supporting the concept that the task is not modulated by feedback from peripheral proprioceptors.

Conclusions

These varying effects may relate to the control mechanisms of the task. It has been noted previously that there may be different neural mechanisms for controlling the task (60). This series of experiments analyzed distinctly different tasks, each of which would be a common component of bipedal locomotion. Using hopping as a control analogue for running, and building on the premise of the spring-loaded inverted pendulum (SLIP) it was noted that there were distinctive differences in the mechanical properties of the leg between the control and anesthetized conditions. Concurrent with this change in stiffness were changes in muscle activation, at least temporally. Studying simple forward gait revealed some kinematic differences between conditions, yet no myologic changes , and gait initiation demonstrated that loss of plantar feedback had only a minimal impact on the various parameters of this task.

As depicted in figure 5-1, the control of the movements measured may have played a significant factor in the response to the loss of feedback, and hence whether or not the changes were detectable by the methods employed. The task of hopping, a rapid motion, would be affected by the mechanical properties of the leg, i.e. the stiffness. The preflexive activity of the muscle, with its onset prior to touchdown, demonstrated that neural feedback is a factor in regulating the mechanical systems.

The SLIP served as a template for the model of hopping, what was defined in this project as a control analogue for running. A template is essentially the simplest pattern that can be used to describe the behavior of a body in motion. Figure 5-2 depicts the modeling hierarchy, and its application within this project.

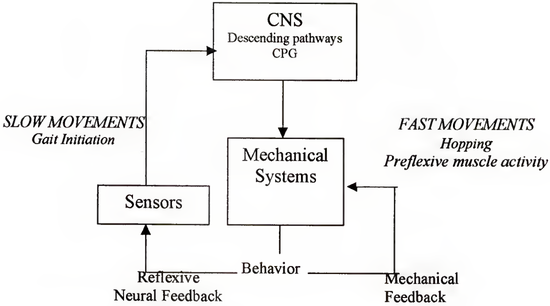

Figure 5-1. Control of actions. A depiction, overly-simplified, of the control of common actions of legged locomotion.

An attempt to integrate the mechanical and neurological processes involved in human locomotion is an essential aspect of understanding the regulation of motion. Subsequent to this knowledge would be the logical and systematic creation of solutions to pathologies in locomotion. Using a SLIP as a model of leg mechanics in the first part of the experiment provided a chance to tests whether this would serve as a valid template for the biological application of the theory. Adding the role of neural input to the template, with its apparent effects on muscle activation, demonstrates the dynamic intercoupling of the nervous and musculoskeletal systems in regulating motion.

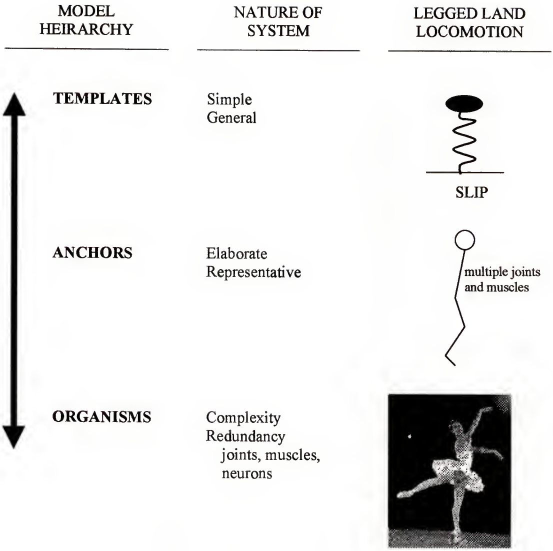

Figure 5-2. Modeling hierarchy for legged locomotion.

The combined phases of this series of experiments point to the Full model for neuromechanical control, depicted previously, as being accurate with regard to the speed of the movement and the role of feedback in regulating the movements (60). Gait initiation, with slower movements than the hopping or the treadmill walking,

demonstrated a negligible effect due to the anesthesia, likely the result of the task being pre-programmed, open loop activity, depending very little on external input to complete the task. In contrast, the rapid dynamic activity of the hopping apparently depends on feedback in order to stabilize the motions.

The conclusions drawn from this project should be to emphasize that mechanical models are insufficient to adequately and accurately describe locomotion. Therefore they are incapable of predicting the forces experienced in motion. Only by including neural pathways, and the role of neural feedback in muscular control, can adequate anchors for terrestrial locomotion be developed.

Future Directions

The results demonstrate that feedback from the plantar aspect of the foot has a role in regulating some aspects of locomotion. Further, the role seems more prominent in rapid activities than it does in more centrally programmed actions. What is now needed is an identification of the receptors that exert the effects this project noted. The decrements in postural stability noted with the loss of plantar feedback may be attributable to the cutaneous mechanoreceptors, but the feedback during the dynamic activities may come from afferents located in the intrinsic musculature, tendons or the multitude of joints located within the foot.

In combination with data showing that people modulate leg stiffness in relation to surface stiffness, and that plantar feedback is an element in regulating leg stiffness, we can reasonably state that adjusting leg stiffness as people move across different surfaces is facilitated by receptors in the foot. Some authors have implicated plantar feedback as

important in minimizing forces, and thus in reducing the likelihood of injuries (103-107). With the knowledge that plantar feedback can affect the mechanical properties of the leg, a forward progression of this line of research should involve determining the specific nature of these changes, the effects that footwear has on plantar feedback, and whether the footwear induces changes similar to what has been observed in this project.

APPENDIX A

INFORMED CONSENT FORM

IRB# 464-1999

Informed Consent to Participate in Research

The University of Florida
Health Science Center
Gainesville, Florida 32610

You are being asked to participate in a research study. This form provides you with information about the study. The Principal Investigator (the person in charge of this research) or his/her representative will also describe this study to you and answer all of your questions. Read the information below and ask questions about anything you don't understand before deciding whether or not to take part. Your participation is entirely voluntary and you can refuse to participate without penalty or loss of benefits to which you are otherwise entitled.

Name of the Subject

Principal Investigator(s) and Telephone Number(s)

Paul Fiolkowski, MA, ATC	(352)-392-0580 x321
Denis Brunt, Ed.D., PT.	(352)-395-0085

Sponsor of the Study
There is no sponsor for this study

What is the purpose of this study?
The purpose of this study is to examine how sensation from the sole of the foot may affect how you begin walking, walk on a treadmill, and hop in place.

What will be done if you take part in this research study?

You will attend the data collection at the Department of Physical Therapy on 1 occasion for approximately 2 hours.

For kinematic data collection (measurement of the movement of your body parts), retroreflective markers (small markers that reflect light directly back to the camera lens) will be placed, using double sided adhesive tape, on the point of the shoulder, hip, lateral side of the knee, and lateral ankle. The video data will be recorded from only the legs, and it will not be possible to identify you from the video data.
The data collection will consist of three experimental protocols for 1) walking kinematics 2) leg stiffness calculation and 3) gait initiation.

The anesthetic (lidocaine) will be injected into both legs, below and behind the outside ankle of both legs.

All tests will be done before the injections, in a barefoot and while wearing running shoes, and then at 5, 20, 35 and 50 minutes after the injection, all in barefoot conditions.

After the anesthetic has taken effect, reflective markers will be taped to your knee ankle, and hip and you will walk on a treadmill at 3 miles per hour, while being filmed. You will walk on the treadmill for 5 minutes. You will then hop on a force plate at a specified frequency (2 times per second) while data are collected. You will do this for 3 trials, each trial lasting 30 seconds with a 1 minute rest between trials. You will then stand on a force plate and then, at a signal from the examiner, start to walk forward, repeating this 10 times. The order of the actions may change.

What are the possible discomforts and risks?
There is the possibility of discomfort with the loss of sensation of the sole of the foot. If this becomes too great the study will be immediately stopped.

There is a risk of soft tissue injury, (bruising) from striking the ground too hard with your foot as you lose sensation in the bottom of your foot.

There is the possibility that loss of the sensation from the sole of the foot may adversely affect your balance while the anesthetic is effective. You will be guarded during all activities to prevent you from falling and will remain in thew lab until the sensation returns to normal.

There is a risk of soreness or infection from the injection of the anesthetic. There is also the risk of an allergic reaction to the anesthetic injection. The risk of using a needle for anesthetic injection include discomfort at the site of puncture; possible bruising and swelling around the puncture site; rarely an infection; and, uncommonly, faintness from the procedure.

If you wish to discuss the information above or any other discomforts you may experience, you may ask questions now or call the Principal Investigator listed on the front page of this form

What are the possible benefits to you or to others?
There are no benefits to you from participation in this study

If you choose to take part in this study, will it cost you anything?
There are no costs to you for being involved in this study.

Will you receive compensation for your participation in this study?

You will receive no compensation for your participation in this study.

What if you are injured because of the study?
If you experience an injury that is directly caused by this study, only

_____professional medical _____professional dental __X__professional consultative care

that you receive at the University of Florida Health Science Center will be provided without charge. However, hospital expenses will have to be paid by you or your insurance provider. No other compensation is offered.

If you do not want to take part in this study, what other options or treatments are available to you?

Participation in this study is entirely voluntary. You are free to refuse to be in the study, and your refusal will not influence current or future health care you receive at this institution.

You have been invited to participate in this research project and you are a student. While students are not the only people being recruited for this study, (subjects are also being recruited for the general healthy adult population of Gainesville) students are being recruited due to their availability to participate during the day. The investigators associated with this project may or may not teach in your college or be associated with courses for which you are enrolled or might be expected to register in the future. Your participation in this study is voluntary and any decision to take part or not to participate will in no way affect your grade or class standing.

If you believe that your participation in this study or your decision to withdraw from or to not participate in this study has improperly affected your grade(s), you should discuss this with the dean of your college or you may contact the IRB office.

How can you withdraw from this research study?

If you wish to stop your participation in this research study for any reason, you should contact:
Paul Fiolkowski at (352) 392-0580 x321 . You are free to withdraw your consent and stop participation in this research study at any time without penalty or loss of benefits to which you are otherwise entitled. Throughout the study, the researchers will notify you of new information that may become available and that might affect your decision to remain in the study.

In addition, if you have any questions regarding your rights as a research subject, you may phone the Institutional Review Board (IRB) office at (352) 846-1494.

How will your privacy and the confidentiality of your research records be protected?

Authorized persons from the University of Florida, the hospital or clinic (if any) involved in this research, and the Institutional Review Board have the legal right to review your research records and will protect the confidentiality of those records to the extent permitted by law. If the research project is sponsored or if it is being conducted under the authority of the United States Food and Drug Administration (FDA), then the sponsor, the sponsor's agent, and the FDA also have the legal right to review your research records. Otherwise, your research records will not be released without your consent unless required by law or a court order.

If the results of this research are published or presented at scientific meetings, your identity will not be disclosed. The only thing that will be filmed is the movement of reflectors taped to your legs. The information about the reflectors movements will be stored directly on the computer. There will be no identifier linking your data to your name. In addition, all data will be kept secured in the Department of Physical Therapy, with only the principal investigator having access to the files. After the research is completed, the data will be kept secured with only the P.I. and his supervisor having access to the files.

Will the researchers benefit from your participation in this study (beyond publishing or presenting the results)?

The researchers will receive no benefit other than publication or presentation of the results.

Signatures

As a representative of this study, I have explained the purpose, the procedures, the benefits, and the risks that are involved in this research study:

____________________________________ ________________

Signature of person obtaining consent Date

You have been informed about this study's purpose, procedures, possible benefits and risks, and you have received a copy of this form. You have been given the opportunity to ask questions before you sign, and you have been told that you can ask other questions at any time. You voluntarily agree to participate in this study. By signing this form, you are not waiving any of your legal rights.

____________________________________ ________________

Signature of Subject Date

____________________________ ________________

Signature of Witness (if available) Date

Appendix B

DATA COLLECTION FORMS

Subject:____________________________________
Height:____________ Weight (kg):______ Age:______ Sex:_______

TESTING TIMES
Familiarization:__________
Control:_______________
1st post-injection:_________
2nd post-injection:_________
3rd post-injection:_________

Physician:____________________
Concentration:_________________
Time of Injection: _______________
Plantar Sensation:_______________

Time sensation begins to return:_________
Time sensation returns to normal:________

GAIT INITIATION
file name:__________________
EMG Leads:__________________

Calibration:__________________

FORCE PLATE:
Calibration:__________________
NOTES:____________________

HOPPING
file name:__________________
EMG Leads:__________________

Calibration:__________________
FORCE PLATE:
Calibration:__________________
VIDEO:
Markers(#/placement):____________

Camera distance (m):____________
NOTES:____________________

TREADMILL
file name:__________________
Markers(#/placement):____________

Camera distance (m):____________

Treadmill Speed:______________

NOTES:____________________

PLANTAR SENSATION
file name:__________________
Romberg____________________

Tactile sensation:______________

Right Foot

Left Foot

Plantar sensation testing:____
Date:______________
Monofilaments__________________ Pressure Algometer__________________

Pressure detection (lbs.)
____

Pressure detection (lbs.)
____

Pressure detection (lbs.)
____

Pressure detection (lbs.)
____

BIBLIOGRAPHY

1. Alexander, R. M. Optimization and gaits in the locomotion of vertebrates. *Physiol Rev* 69(4):1199-227, 1989.

2. Alexander, R. M. Elastic mechanisms in primate locomotion. *Z Morphol Anthropol* 78(3):315-20, 1991.

3. Alexander, R. M. A model of bipedal locomotion on compliant legs. *Philos Trans R Soc Lond B Biol Sci* 338(1284):189-98, 1992.

4. Anderson, A. M., and R. P. Croft. Reliability of Semmes Weinstein monofilament and ballpoint sensory testing, and voluntary muscle testing in Bangladesh. *Lepr Rev* 70(3):305-13, 1999.

5. Aniss, A. M., H. C. Diener, J. Hore, S. C. Gandevia, and D. Burke. Behavior of human muscle receptors when reliant on proprioceptive feedback during standing. *J Neurophysiol* 64(2):661-70, 1990.

6. Antonaci, F., T. Sand, and G. A. Lucas. Pressure algometry in healthy subjects: inter-examiner variability. *Scand J Rehabil Med* 30(1):3-8, 1998.

7. Arampatzis, A., G. P. Bruggemann, and V. Metzler. The effect of speed on leg stiffness and joint kinetics in human running. *J Biomech* 32(12):1349-53, 1999.

8. Barrett, R. S., R. J. Neal, and L. J. Roberts. The dynamic loading response of surfaces encountered in beach running. *J Sci Med Sport* 1(1):1-11, 1998.

9. Bell-Krotoski, J., and E. Tomancik. The repeatability of testing with Semmes-Weinstein monofilaments. *J Hand Surg [Am]* 12(1):155-61, 1987.

10. Biewener, A. A., and R. Blickhan. Kangaroo rat locomotion: Design for elastic energy storage or acceleration? *J Exp Biol* 140:243-55, 1988.

11. Blickhan, R. The spring mass model for running and hopping. *J Biomech* 22(11/12):1217-1227, 1989.

12. Bloem, B. R., J. H. Allum, M. G. Carpenter, and F. Honegger. Is lower leg proprioception essential for triggering human automatic postural responses? *Exp Brain Res* 130(3):375-91, 2000.

13. Bolton, C. F., R. K. Winkelmann, and P. J. Dyck. A quantitative study of Meissner's corpuscles in man. *Neurology* 16(1):1-9, 1966.

14. Bosco, C., R. Saggini, and A. Viru. The influence of different floor stiffness on mechanical efficiency of leg extensor muscle. *Ergonomics* 40(6):670-9, 1997.

15. Breniere, Y., and M. Do. When and how does steady state gait movement induced from upright posture begin? *J Biomech* 19:1035-1040, 1986.

16. Breniere, Y., and M. Do. Control of gait initiation. *Journal of Motor Behavior* 23(4):235-240, 1991.

17. Breniere, Y., M. Do, and S. Bouisset. Are dynamic phenomena prior to stepping essential? *Journal of Motor Behavior* 19(1):62-76, 1987.

18. Brown, I., and G. Loeb. A reductionist approach to creating and using neuromusculoskeletal models. In: *Biomechanics and Neural Control of Movement*. J. Winters and P. Crago (Eds.). New York: Springer-Verlag, 1999.

19. Brunt, D., M. J. Lafferty, A. McKeon, B. Goode, C. Mulhausen, and P. Polk. Invariant characteristics of gait initiation. *Am J Phys Med Rehabil* 70(4):206-12, 1991.

20. Brunt, D., S. M. Liu, M. Trimble, J. Bauer, and M. Short. Principles underlying the organization of movement initiation from quiet stance. *Gait Posture* 10(2):121-8, 1999.

21. Brunt, D., M. Short, M. Trimble, and S. Liu. Control strategies for initiation of human gait are influenced by accuracy constraints. *Neuroscience Letters* 285(3):228-230, 2000.

22. Burke, D., S. C. Gandevia, and G. Macefield. Responses to passive movement of receptors in joint, skin and muscle of the human hand. *J Physiol (Lond)* 402:347-61, 1988.

23. Carter, R. R., P. E. Crago, and M. W. Keith. Stiffness regulation by reflex action in the normal human hand. *Journal of Neurophysiology* 64(1):105-118, 1990.

24. Cavagna, G. A. Force platforms as ergometer. *Journal of Applied Physiology* 39(1):174-179, 1975.

25. Cavagna, G. A., P. Franzetti, N. C. Heglund, and P. Willems. The determinants of the step frequency in running, trotting and hopping in man and other vertebrates. *J Physiol (Lond)* 399:81-92, 1988.

26. Cavagna, G. A., M. Mazzanti, N. C. Heglund, and G. Citterio. Storage and release of mechanical energy by active muscle: a non- elastic mechanism? *J Exp Biol* 115:79-87, 1985.

27. Cavanagh, P. R. (Ed.) *Biomechanics of Long Distance Running.* Champaign, IL: Human Kinetics Publishers, 1992.

28. Cavanagh, P. R., G. G. Simoneau, and J. S. Ulbrecht. Ulceration, unsteadiness, and uncertainty: the biomechanical consequences of diabetes mellitus. *J Biomech* 26(Suppl 1):23-40, 1993.

29. Chang, H., and D. E. Krebs. Dynamic balance control in elders: gait initiation assessment as a screening tool. *Arch Phys Med Rehabil* 80(5):490-4, 1999.

30. Clark, F. J., P. Grigg, and J. W. Chapin. The contribution of articular receptors to proprioception with the fingers in humans. *J Neurophysiol* 61(1):186-93, 1989.

31. Clark, F. J., K. W. Horch, S. M. Bach, and G. F. Larson. Contributions of cutaneous and joint receptors to static knee-position sense in man. *Journal of Physiology* 42(3):877-888, 1979.

32. Cole, K. J., B. J. Daley, and R. A. Brand. The sensitivity of joint afferents to knee translation. *Sportverletz Sportschaden* 10(2):27-31, 1996.

33. Conway, B., D. Scott, J. Riddell, and M. Hadian. Effects of plantar nerve stimulation on the transmision of late flexion reflexes in the spinal cat. *J Physiol (Lond)* 496:857-871, 1994.

34. Crenna, P., and C. Frigo. A motor programme for the initiation of forward-oriented movements in humans. *J Physiol (Lond)* 437:635-53, 1991.

35. Czerniecki, J. M., R. M. Harrington, C. R. Wyss, B. J. Sangeorzan, and F.A.D. Matsen. The effects of age and peripheral vascular disease on the circulatory and mechanical response of skin to loading. *Am J Phys Med Rehabil* 69(6):302-6, 1990.

36. Dalleau, G., A. Belli, M. Bourdin, and J. R. Lacour. The spring-mass model and the energy cost of treadmill running. *Eur J Appl Physiol* 77(3):257-63, 1998.

37. De Wit, B., D. De Clercq, and P. Aerts. Biomechanical analysis of the stance phase during barefoot and shod running. *J Biomech* 33(3):269-78, 2000.

38. Dietz, V., G. Colombo, L. Jensen, and L. Baumgartner. Locomotor capacity of spinal cord in paraplegic patients [see comments]. *Ann Neurol* 37(5):574-82, 1995.

39. Dietz, V., and J. Duysens. Significance of load receptor input during locomotion: a review. *Gait Posture* 11(2):102-10, 2000.

40. Dietz, V., D. Schmidtbleicher, and J. Noth. Neuronal Mechanisms of Human locomotion. *Journal of Neurophysiology* 42(5):1212-1222, 1979.

41. Dingwell, J. B., J. S. Ulbrecht, J. Boch, M. B. Becker, J. T. O'Gorman, and P. R. Cavanagh. Neuropathic gait shows only trends towards increased variability of sagittal plane kinematics during treadmill locomotion. *Gait Posture* 10(1):21-9, 1999.

42. Do, M. C., B. Bussel, and Y. Breniere. Influence of plantar cutaneous afferents on early compensatory reactions to forward fall. *Experimental Brain Research* 79(2):319-324, 1990.

43. Dodson, M. J., A. W. Goodwin, A. S. Browning, and H. M. Gehring. Peripheral neural mechanisms determining the orientation of cylinders grasped by the digits. *J Neurosci* 18(1):521-30, 1998.

44. Duysens, J., F. Clarac, and H. cruse. Load regulating mechanisms in gait and posture. *Physiological Reviews* 80(1):83-133, 2000.

45. Duysens, J., and K. G. Pearson. The role of cutaneous afferents from the distal hindlimb in the regulation of the step cycle of thalamic cats. *Exp Brain Res* 24:245-55, 1976.

46. Duysens, J., and H. W. Van de Crommert. Neural control of locomotion; The central pattern generator from cats to humans. *Gait Posture* 7(2):131-141, 1998.

47. Farley, C. T., R. Blickhan, J. Saito, and C. R. Taylor. Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits. *Journal of Applied Physiology* 71(6):2127-2132, 1991.

48. Farley, C. T., and D. P. Ferris (Eds.) *Biomechanics of Walking and Running: Center of Mass Movements to Muscle Action*, Vol. 26. Baltimore: Williams and Wilkins, 1998.

49. Farley, C. T., J. Glasheen, and T. A. McMahon. Running springs: Speed and Animal size. *Journal of Experimental Biology* 185:71-86, 1993.

50. Farley, C. T., and O. Gonzalez. Leg stiffness and stride frequency in human running. *J Biomech* 29(2):181-186, 1996.

51. Farley, C. T., H. H. Houdjick, C. V. Strein, and M. Louie. Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. *Journal of Applied Physiology* 85(3):1044-1055, 1998.

52. Farley, C. T., and D. C. Morgenroth. Leg stiffness primarily depends on ankle stiffness during human hopping. *J Biomech* 32(3):267-73, 1999.

53. Fedak, M. A., N. C. Heglund, and C. R. Taylor. Energetics and mechanics of terrestrial locomotion. II. Kinetic energy changes of the limbs and body as a function of speed and body size in birds and mammals. *J Exp Biol* 97:23-40, 1982.

54. Feehery, R. V., Jr. The biomechanics of running on different surfaces. *Clin Podiatr Med Surg* 3(4):649-59, 1986.

55. Ferrell, W. R., and A. Smith. The effect of loading on position sense at the proximal interphalangeal joint of the human index finger. *J Physiol (Lond)* 418:145-61, 1989.

56. Ferris, D. P., and C. T. Farley. Interaction of leg stiffness and surfaces stiffness during human hopping. *J Appl Physiol* 82(1):15-22; discussion 13-4, 1997.

57. Ferris, D. P., K. Liang, and C. T. Farley. Runners adjust leg stiffness for their first step on a new running surface. *J Biomech* 32(8):787-94, 1999.

58. Ferris, D. P., M. Louie, and C. T. Farley. Running in the real world: Adjusting leg stiffness for different surfaces. *Proc R Soc Lond B Biol Sci* 265(1400):989-94, 1998.

59. Full, R., A. Yamauchi, and D. Jindrich. Maximum single leg force production: Cockroaches righting on photoelastic gelatin. *J Exp Biol* 198(Pt):2441-52, 1995.

60. Full, R. J., and D. E. Koditschek. Templates and anchors: Neuromechanical hypotheses of legged locomotion on land. *J Exp Biol* 202 Pt 23:3325-32, 1999.

61. Full, R. J., and M. S. Tu. Mechanics of a rapid running insect: Two-, four- and six-legged locomotion. *J Exp Biol* 156:215-31, 1991.

62. Grigg, P. Peripheral neural mechanisms and proprioception. *Journal of Sport Rehabilitation* 3:2-17, 1994.

63. Hamill, J., and K. Knutzen. *Biomechanical Basis of Human Movement*. Baltimore: Williams and Wilkins, 1995.

64. Hansen, P. *Neurology of Walking*. Gainesville, FL: 2000.

65. Heglund, N. C., G. A. Cavagna, and C. R. Taylor. Energetics and mechanics of terrestrial locomotion. III. Energy changes of the centre of mass as a function of speed and body size in birds and mammals. *J Exp Biol* 97:41-56, 1982.

66. Heise, G. D., and P. E. Martin. "Leg spring" characteristics and the aerobic demand of running. *Med Sci Sports Exerc* 30(5):750-4, 1998.

67. Herman, R., T. Cook, B. Cozzens, and W. Freedman. Control of postural reactions in man: the initiation of gait. In: *Control of Posture and Locomotion*. R. Stein, K. Pearson, R. Smith, and J. Redford (Eds.). New York: Plenum Press, 1973, pp. 363-388.

68. Hesse, S., M. Konrad, and D. Uhlenbrock. Treadmill walking with partial body weight support versus floor walking in hemiparetic subjects. *Arch Phys Med Rehabil* 80(4):421-7, 1999.

69. Hogervorst, T. Knee ligaments and proprioception [letter; comment]. *Acta Orthop Scand* 68(2):183-5, 1997.

70. Iles, J. F. Evidence for cutaneous and corticospinal modulation of presynaptic inhibition of Ia afferents from the human lower limb. *J Physiol (Lond)* 491(Pt 1):197-207, 1996.

71. Johansson, R. S., and G. Westling. Roles of glabrous skin receptors and sensorimotor memory in automatic control of precision grip when lifting rougher or more slippery objects. *Exp Brain Res* 56(3):550-64, 1984.

72. Jones, L. A. Peripheral mechanisms of touch and proprioception. *Canadian Journal of Physiology and Pharmacology* 72:484-487, 1994.

73. Kavounoudias, A., J. C. Gilhodes, R. Roll, and J. P. Roll. From balance regulation to body orientation: two goals for muscle proprioceptive information processing? *Exp Brain Res* 124(1):80-8, 1999.

74. Kavounoudias, A., R. Roll, and J. P. Roll. The plantar sole is a 'dynamometric map' for human balance control. *Neuroreport* 9(14):3247-52, 1998.

75. Khalsa, P. S., C. Zhang, and Y. X. Qin. Encoding of location and intensity of noxious indentation into rat skin by spatial populations of cutaneous mechano-nociceptors. *J Neurophysiol* 83(5):3049-61, 2000.

76. Lafortune, M. A., E. M. Hennig, and M. J. Lake. Dominant role of interface over knee angle for cushioning impact loading and regulating initial leg stiffness. *J Biomech* 29(12):1523-9, 1996.

77. Langman, V. A., T. J. Roberts, J. Black, et al. Moving cheaply: Energetics of walking in the African elephant. *J Exp Biol* 198(Pt 3):629-32, 1995.

78. Leem, J. W., W. D. Willis, S. C. Weller, and J. M. Chung. Differential activation and classification of cutaneous afferents in the rat. *J Neurophysiol* 70(6):2411-24, 1993.

79. List, T., M. Helkimo, and G. Falk. Reliability and validity of a pressure threshold meter in recording tenderness in the masseter muscle and the anterior temporalis muscle. *Cranio* 7(3):223-9, 1989.

80. List, T., M. Helkimo, and R. Karlsson. Influence of pressure rates on the reliability of a pressure threshold meter. *J Craniomandib Disord* 5(3):173-8, 1991.

81. Lord, S. R., G. A. Caplan, R. Colagiuri, S. Colagiuri, and J. A. Ward. Sensori-motor function in older persons with diabetes. *Diabet Med* 10(7):614-8, 1993.

82. Lord, S. R., P. N. Sambrook, C. Gilbert, et al. Postural stability, falls and fractures in the elderly: results from the Dubbo Osteoporosis Epidemiology Study. *Med J Aust* 160(11):684-5, 688-91, 1994.

83. Lord, S. R., J. A. Ward, P. Williams, and K. J. Anstey. Physiological factors associated with falls in older community-dwelling women. *J Am Geriatr Soc* 42(10):1110-7, 1994.

84. Lundberg, A., K. Malmgren, and E. D. Schomburg. Cutaneous facilitation of transmission in reflex pathways from Ib afferents to motoneurones. *J Physiol (Lond)* 265(3):763-80, 1977.

85. MacDermid, J. C., J. F. Kramer, and J. H. Roth. Decision making in detecting abnormal Semmes-Weinstein monofilament thresholds in carpal tunnel syndrome. *J Hand Ther* 7(3):158-62, 1994.

86. Macefield, V. G., C. Hager-Ross, and R. S. Johansson. Control of grip force during restraint of an object held between finger and thumb: responses of cutaneous afferents from the digits. *Exp Brain Res* 108(1):155-71, 1996.

87. Magnusson, M., H. Enbom, R. Johansson, and I. Pyykko. Significance of pressor input from the human feet in anterior-posterior posture control. *Acta Otolaryngolica* 110:182-188, 1990.

88. Margaria, R., and G. A. Cavagna. Human locomotion on the moon surface. *Riv Med Aeronaut Spaz* 30(4):629-44, 1967.

89. McMahon, T. A. The role of compliance in mammalian running gaits. *J Exp Biol* 115:263-82, 1985.

90. McMahon, T. A., and G. C. Cheng. The mechanics of running: How does stiffness couple with speed? *J Biomech* 23(s1):65-78, 1990.

91. McMahon, T. A., G. Valiant, and E. C. Frederick. Groucho running. *Journal of Applied Physiology* 62(6):2326-2337, 1987.

92. McNair, P. J., and R. N. Marshall. Kinetic and kinematic parameters associated with runing in different shoes. *British Journal of Sports Medicine* 28(4):2560-2660, 1994.

93. Mittelstaedt, H. Somatic graviception. *Biol Psychol* 42(1-2):53-74, 1996.

94. Moberg, E. The role of cutaneous afferents in position sense, kinaesthesia, and motor function of the hand. *Brain* 106(Pt 1):1-19, 1983.

95. Nielsen, J., N. Petersen, and B. Fedirchuk. Evidence suggesting a transcortical pathway from cutaneous foot afferents to tibialis anterior motoneurones in man. *J Physiol (Lond)* 501(Pt 2):473-84, 1997.

96. Nielsen, J., T. Sinkjaer, E. Toft, and Y. Kagamahira. Segmental reflexes and ankle joint stiffness during co-contraction of antagonist ankle muscles in man. *Experimental Brain research* 102:350-358, 1994.

97. Nigg, B. M., and Bobbert. On the potential of various approaches in load analysis to reduce the frequency of sports injuries. *Journal of Biomechanics* 23(S1):3-12, 1990.

98. Nigg, B. M., G. K. Cole, and G. P. Bruggeman. Impact forces during heel-toe running. *Journal of Applied Biomechanics* 11:407-442, 1995.

99. Nussbaum, E. L., and L. Downes. Reliability of clinical pressure-pain algometric measurements obtained on consecutive days. *Phys Ther* 78(2):160-9, 1998.

100. Obusek, J. P., K. G. Holt, and R. M. Rosenstein. The hybrid mass-spring pendulum model of human leg swinging: stiffness in the control of cycle period. *Biol Cybern* 73(2):139-47, 1995.

101. Ohrbach, R., and E. N. Gale. Pressure pain thresholds, clinical assessment, and differential diagnosis: reliability and validity in patients with myogenic pain. *Pain* 39(2):157-69, 1989.

102. Patel, M. R., and L. Bassini. A comparison of five tests for determining hand sensibility. *J Reconstr Microsurg* 15(7):523-6, 1999.

103. Polcyn, A. F., L. A. Lipsitz, D. C. Kerrigan, and J. J. Collins. Age-related changes in the initiation of gait: degradation of central mechanisms for momentum generation. *Arch Phys Med Rehabil* 79(12):1582-9, 1998.

104. Prete, Z. D., and P. Grigg. Responses of rapidly adapting afferent neurons to dynamic stretch of rat hairy skin. *J Neurophysiol* 80(2):745-54, 1998.

105. Pyykko, I., P. Jantti, and H. Aalto. Postural control in elderly subjects. *Age Ageing* 19(3):215-21, 1990.

106. Reeves, J. L., B. Jaeger, and S. B. Graff-Radford. Reliability of the pressure algometer as a measure of myofascial trigger point sensitivity. *Pain* 24(3):313-21, 1986.

107. Refshauge, K. M., and R. C. Fitzpatrick. Perception of movement at the human ankle: effects of leg position. *J Physiol (Lond)* 488(Pt 1):243-8, 1995.

108. Robbins, S., E. Waked, P. Allard, J. McClaran, and N. Krouglicof. Foot position awareness in younger and older men: the influence of footwear sole properties. *J Am Geriatr Soc* 45(1):61-66, 1997.

109. Robbins, S. E., and G. J. Gouw. Athletic footwear and chronic overloading: A brief review. *Sports Medicine* 9(2):76-85, 1990.

110. Robbins, S. E., G. J. Gouw, and A. M. Hanna. Running related injury prevention through impact-moderating behavior. *Medicine and Science in Sports and Exercise* 21(2):130-139, 1989.

111. Robbins, S. E., and A. M. Hanna. Running related injury prevention through barefoot adaptation. *Medicine and Science in Sports and Exercise* 19(2):148-156, 1987.

112. Robbins, S. E., A. M. Hanna, and G. J. Gouw. Overload protection: Avoidance response to heavy plantar surface loading. *Medicine and Science in Sports and Exercise* 20(1):85-92, 1988.

113. Robbins, S. E., A. M. Hanna, and L. A. Jones. Sensory attenuation induced by modern athletic footwear. *Journal of Testing and Evaluation* 16(4):422-426, 1988.

114. Robbins, S. E., E. Waked, and J. McClaran. Proprioception and stability: Foot position awareness as a function of age and footwear. *Age and Ageing* 24:67-72, 1995.

115. Rossi, A., and B. Decchi. Flexibility of lower limb reflex responses to painful cutaneous stimulation in standing humans: Evidence of load-dependent modulation. *J Physiol (Lond)* 481(Pt 2):521-32, 1994.

116. Rossi, A., A. Zalaffi, and B. Decchi. Interaction of nociceptive and non-nociceptive cutaneous afferents from foot sole in common reflex pathways to tibialis anterior motoneurones in humans. *Brain Research* 714:76-86, 1996.

117. Schaible, H. G., R. F. Schmidt, and W. D. Willis. Convergent inputs from articular, cutaneous and muscle receptors onto ascending tract cells in the cat spinal cord. *Exp Brain Res* 66(3):479-88, 1987.

118. Schimrigk, K., and H. Ruttinger. The touch corpuscles of the plantar surface of the big toe. Histological and histometrical investigations with respect to age. *Eur Neurol* 19(1):49-60, 1980.

119. Schultz, R. A., D. C. Miller, C. S. Kerr, and L. Micheli. Mechanoreceptors in human cruciate ligaments. A histological study. *J Bone Joint Surg [Am]* 66(7):1072-6, 1984.

120. Sherrington, C. *The Integrative Action of the Nervous System*. New Haven: Yale University Press, 1911.

121. Sillar, K. T., and A. Roberts. A neuronal mechanism for sensory gating during locomotion in a vertebrate. *Nature* 331(6153):262-5, 1988.

122. Smith, J. L., V. R. Edgerton, E. Eldred, and R. F. Zernicke. The chronic spinalized cat: a model for neuromuscular plasticity. *Birth Defects Orig Artic Ser* 19(4):357-73, 1983.

123. Stelmach, G. E., and C. J. Worringham. Sensorimotor deficits related to postural stability. Implications for falling in the elderly. *Clin Geriatr Med* 1(3):679-94, 1985.

124. Stephens, M. J., and J. F. Yang. Loading during the stance phase of walking in humans increases the extensor EMG amplitude but does not change the duration of the step cycle. *Exp Brain Res* 124(3):363-70, 1999.

125. Thoumie, P., and M. C. Do. Changes in motor activity and biomechanics during balance recovery following cutaneous and muscular deafferentation. *Exp Brain Res* 110(2):289-97, 1996.

126. Thurner, J., M. Marcacci, and L. Felli. [Articular capsule receptors, articular nerves, intra-articular multipolar ganglion cells and their possible functional importance (author's transl)]. *Z Orthop Ihre Grenzgeb* 119(3):301-5, 1981.

127. Vallbo, A. B., and R. S. Johansson. Properties of cutaneous mechanoreceptors in the human hand related to touch sensation. *Hum Neurobiol* 3(1):3-14, 1984.

128. van Brakel, W. H., I. B. Khawas, K. S. Gurung, C. M. Kets, M. E. van Leerdam, and W. Drever. Intra- and inter-tester reliability of sensibility testing in leprosy. *Int J Lepr Other Mycobact Dis* 64(3):287-98, 1996.

129. Van de Crommert, H. W., T. Mulder, and J. Duysens. Neural control of locomotion: sensory control of the central pattern generator and its relation to treadmill training. *Gait Posture* 7(3):251-263, 1998.

130. van Deursen, R. W., and G. G. Simoneau. Foot and ankle sensory neuropathy, proprioception, and postural stability. *J Orthop Sports Phys Ther* 29(12):718-26, 1999.

131. Wagner, H., and R. Blickhan. Stabilizing function of skeletal muscles: an analytical investigation. *J Theor Biol* 199(2):163-79, 1999.

132. Williams, W. J. A systems-oriented evaluation of the role of joint receptors and other afferents in position and motion sense. *Crit Rev Biomed Eng* 7(1):23-77, 1981.

133. Winter, D. A. *Biomechanics and Motor Control of Human Movement.* New York: John Wiley and Sons, Inc., 1990.

BIOGRAPHICAL SKETCH

Paul Fiolkowski graduated from Miami University in 1985,earned a Masters of Arts in 1993 from Glassboro State College and will work as a postdoctoral research associate in the Department of Physical Therapy at the University of Florida.

SÍN E

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

Stephen L. Dodd
Associate Professor of Exercise and Sport Sciences

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

MaryBeth Horodyski
Associate Professor of Exercise and Sport Sciences

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

Denis Brunt
Associate Professor of Physical Therapy

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

Mark Trimble
Associate Professor of Physical Therapy

This dissertation was submitted to the Graduate Faculty of the College of health and Human Performance and to the Graduate School and was accepted as partial fulfillment of the requirements for the degree of Doctor of Philosophy.

December, 2000

Dean, College of Health And Human Performance

Dean, Graduate School

General Audience Abstract

THE ROLE OF PLANTAR FEEDBACK IN THE REGULATION OF HOPPING, WALKING AND GAIT INITIATION: A NEUROMECHANICAL APPROACH TO LEGGED LOCOMOTION

Paul Fiolkowski, MA, ATC, CSCS
352.395.0085
Exercise and Sport Sciences
Stephen L. Dodd, Ph.D.
Doctor of Philosophy
Fall, 2000

When people run or walk over different, they are capable of adapting their leg to different terrain. Optimal gait requires adjustment of leg stiffness, inversely proportional to the stiffness of the surface. To date, there has been no attempt to identify the mechanisms by which these adjustments are possible.

This project consisted of three experiments to determine the role that plantar feedback has in regulating leg stiffness, forward gait and gait initiation. lidocaine injections were used to block the tibial nerve at the level of the ankle, eliminating feedback from the sole of the foot. The results indicate that plantar feedback is important in regulating leg stiffness, and exerts an effect on muscle activation of the quadriceps and gastrocnemius. There was less of an effect noted when walking on a treadmill, and in gait initiation, there appears to be little effect caused by the loss of plantar sensation.